

FORCES IN THE JOINTS OF THE LOWER LIMB
BEFORE AND AFTER HIP ARTHROPLASTY

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DECLARATION

This study was carried out whilst the author was involved in an M.R.C. sponsored study of loads in the lower limb joints after a variety of operative procedures. This work was under the supervision of Professor D.L. Hamblen of the Department of Orthopaedics, University of Glasgow, and Professor J.P. Paul of the Bioengineering Unit, University of Strathclyde.

The gait analysis tests and data analysis were performed in conjunction with Dr. T.R.M. Brown of the Bioengineering Unit of the University of Strathclyde. The computer programmes used in these analyses were modified from existing programmes by Dr. Brown.

The concept of studying anatomical changes at the hip occasioned by surgery, the acquisition of the clinical data, the radiological measurements, the interpretation of the biomechanical data with regard to the clinical findings and the statistical analyses represent the author's individual contribution to the study.

ABSTRACT

The work in this thesis was undertaken to investigate the possible biomechanical basis for the clinical observation of degenerative changes occurring in the knees of patients with rheumatoid arthritis who had undergone hip joint replacement. To investigate the effect of pathological and reconstructed hip anatomy upon lower limb joint loading, a group of patients who had undergone excision arthroplasty of the hip joint for failed hip replacement was studied in addition to patients with degenerative hip disease before and after total hip replacement using two different types of hip arthroplasty.

The development of gait analysis has been reviewed together with the use of this technique in the investigation of hip disease and joint replacement.

Clinical assessments were made using the Harris evaluation system and biomechanical tests were performed using a television-force-plate-computer system. This system allows a three-dimensional assessment of the movement of the lower limbs during walking and correlates this information with ground reaction force to the subject obtained from the force plate. Synchronisation of this information allows an estimate of hip and knee joint forces to be made using free body analysis and a mathematical model of the disposition of the muscle groups about the joints and their phasic activity.

To allow for the effects of the abnormal anatomy in the three groups measurements were taken from x-rays and used to "tailor" the muscle model for each individual patient.

Linear regression analyses and correlations were performed upon the anatomical data in an attempt to identify those factors of greatest importance in determining joint loads.

The results obtained in this study have been discussed in the light of previous work defining features of the gait pattern for each group and the possible influence of prosthetic design and surgical technique upon the lower limb joint loads.

The major conclusions are that all three groups of patients exhibit a similar gait pattern involving abnormalities of movement and loading at both hip and knee joints. This appears to result from hip muscle dysfunction. It is hypothesised that the limitation of extension and internal rotation at the affected hip joint is an important factor with regard to the movement and loads seen at the knee joints.

NOTATION

The symbol \circ represents a suffix

A	Ankle joint centre
a	Distance between the anterior superior iliac spines, measured from the subject
a'	Distance CB on radiograph i.e. the distance between the anterior iliac spines
B	Anterior superior iliac spine marker on side distant from camera
BW	Body weight
b	Distance between B and tail marker
C	Anterior superior iliac spine marker nearest to camera
c	Distance between C and tail marker
F	Foot
FD	Design of femoral component used
FGo	Component of ground reaction force
GS	Girdlestone excision arthroplasty
GT	Greater trochanter
GTZ	Horizontal distance between greater trochanter and hip joint centre
H	Hip joint centre
HJF	Resultant hip joint force
HJR	Hip joint replacement
H.S.	Heel strike
h	Height of camera above reference axis
Io	Insertion of muscle group

JKy	Compartmental joint force at knee
K	Knee joint centre
KJF	Resultant knee joint force
Labd	Length of abductor muscle lever arm
LLo	Left Leg
LT	Lesser trochanter
MHo	Moment about designated axis at hip joint
MKo	Moment about designated axis at knee joint
NL	Neck length
Oo	Origin of muscle group or ligament
Po	Tension in muscle group or ligament
Ro	Camera distance from reference axes
RLo	Right leg
S	Shank
T	Thigh
T	Tail marker
TFC	Sum of the force components along the tibial Y axis of the external force system, the active muscles and the cruciate ligaments acting at the knee
T.O.	Toe off
X Y Z	Axis system centred on joint indicated by prefix
x y z	Coordinates of a point in the axis system
x' y' z'	Apparent coordinates of a joint before parallax correction
y	Vertical distance of H from line CB on radiographs

- z Projection of H on to line CB measured from
nearest anterior superior iliac spine on
radiographs
- Yc $y(\text{affected or replaced hip}) - y(\text{healthy hip})$
-
- $y(\text{healthy hip})$
- Zc $z(\text{affected or replaced hip}) - z(\text{healthy hip})$
-
- $z(\text{healthy hip})$

STATISTICS

All means are presented \pm two standard deviations
of the mean.

CHAPTER 1

REVIEW OF THE RELEVANT LITERATURE

- I INTRODUCTION
- II THE DEVELOPMENT OF HIP JOINT REPLACEMENT SURGERY
- III LOCOMOTION STUDIES
 - 1. History
 - 2. Instrumented Implants
 - 3. Other Approaches to the Study of Gait
 - 4. The Need for Computers
- IV MUSCLE FUNCTION DURING GAIT
 - 1. Muscle Mechanics
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- VI STUDIES OF GAIT IN DEGENERATIVE HIP DISEASE AND ARTHROPLASTY OF THE HIP
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 - a) Time and Distance Parameters of Gait
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 - i) Ground Reaction Forces
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- VII THE INFLUENCE OF ANATOMICAL FACTORS AT THE HIP UPON GAIT
- VIII STUDIES OF GAIT AFTER RESURFACING ARTHROPLASTY OF THE HIP
- IX STUDIES OF GAIT AFTER EXCISION ARTHROPLASTY OF THE HIP JOINT

1.I INTRODUCTION

The problem of the management of degenerative disease of the hip joint has stimulated the development of surgical procedures at the joint culminating in the successful form of total hip replacement in use today (Chapter 2.1). However, like any other surgical procedure, total hip arthroplasty is not free of problems. The most important of these are infection, loosening, dislocation and component failure; that is they are mainly mechanical in nature. Recently clinical observations have suggested that hip arthroplasty may be followed by the development of degenerative changes in other joints. A total of 15,300 hip replacements were performed in England and Wales during 1978. With the generally accepted figure of 90% successful procedures it is evident that although the failure rate is low the problem of failure is significant in terms of patient numbers. When it is considered that the management of these failures is costly in terms of the facilities used and operative time in the presence of an ever-present demand for primary surgery, then the need for a more complete understanding of hip joint function in order to avoid failure is evident.

Walking is the principal expression of lower limb function and its study is therefore pertinent to an understanding of hip joint function in health and disease. The development of the methods used for the analysis of gait are summarised (1.III).

The importance of understanding the role of muscle function during locomotion and the methods used to overcome the bioengineering problems presented by muscular arrangements are discussed in 1.IV, and the evidence linking joint loading to degenerative changes in joints is reviewed in 1.V.

There are many parameters of gait which can be measured and the value of these in the study of hip joint disease and joint replacement is discussed in 1.VI together with a critical review of the relevant literature.

The disturbance of anatomy at the hip joint which accompanies hip joint disease and joint replacement changes the mechanics of the hip joint and studies concerning its influence upon the parameters of gait are reported in 1.VII.

Other surgical procedures are used for the management of hip disease and the work relating to the influence of two of these, resurfacing arthroplasty of the hip, and excision arthroplasty, are reviewed (1.VIII and 1.IX).

1.II THE DEVELOPMENT OF HIP JOINT REPLACEMENT SURGERY

The management of the patient with a diseased or damaged hip joint has presented surgeons with a problem for centuries. Surgical remedies have followed one of three paths. The first of these is excision of the head and neck of the femur, proposed in 1770 by Charles White but first carried out by Anthony White at the Westminster Hospital in 1822 for what was probably a dislocated hip. Syme published his "Treatise on the excision of diseased joints" as an alternative to amputation in 1861. However, the mortality of this procedure was in excess of 50% and this led Hugh Owen Thomas to advocate a non-surgical approach to the problem in the late 19th century. With the advent of better antiseptic and anaesthetic techniques G.R. Girdlestone was able to adopt this approach for the management of the tuberculous hip joint in 1924, and later extended it to the treatment of degenerative hip disease (1945). The main problem with this procedure is the resulting instability of the hip joint and its use is now almost entirely confined to the management of the infected total hip replacement (vide infra).

The second path to be followed was that of performing intertrochanteric osteotomy for the correction of flexion deformities. John Rhea Barton performed this procedure in Philadelphia in 1826 and obtained good results by virtue of the reaction of a pseudarthrosis. Osteotomy has advanced since then with contributions from T.P. McMurray in Liverpool (1935) and Pauwels (1935) in Germany. It still has a very

important role in the management of hip disease and is used widely in continental Europe.

The final path taken resulted from the occurrence of stiffness following excision arthroplasty. Verneuil in France (1860) attempted to prevent this by interposing muscle and others (Colonna, 1932) used other tissues and even gold foil (Robert Jones, 1908) with limited success. Smith-Petersen working in Boston in 1923 felt that congruous surfaces were needed to restore movement and maintain stability and noted that a fragment of glass removed from a patient after one year was surrounded by a synovial sac. He therefore used a glass cup as an interposition with the intention that it should act as a mould which would be removed later. Unfortunately the glass moulds broke but eventually a vitallium cup was used and found to be satisfactory.

Gluck in Berlin in 1890 was probably the first to report upon the use of ivory ball and socket joints at the hip, and other joints, in the treatment of war wounds. Unfortunately all of his joints were extruded (Jones, 1908). Hey Groves (1922) again used an ivory femoral head replacement in an ingenious procedure to equalise leg lengths and treat non-union of a fracture of the femoral neck in a man of 41 years. He was more fortunate than Gluck and when Hey Groves reported this case the patient had had four years of active use of the limb.

Hey Groves' procedure bears a striking resemblance to that of the Judet Brothers who re-introduced the

concept of femoral head replacement in 1946. They held the view that osteoarthritis of the hip was a disease of the femoral head and did not feel that cup arthroplasty could be expected to give consistently good results for this reason. Unfortunately, owing to the situation in France after World War II which resulted in the limited availability of materials they had to construct their prosthesis from acrylic resin. This prosthesis had gained wide international acceptance before the inadequacies of the brittle material were appreciated. Indeed, because of mechanical failures of the stem, the design, rather than the material was changed. Later attempts using nylon and steel failed because of adverse tissue reaction and loosening respectively. It was ultimately realised that both the design and the material of the Judet prosthesis were unsatisfactory.

However, the work of Smith-Petersen and the Judet Brothers stimulated further interest in the field of hip joint arthroplasty. Wiles attempted the total replacement of six hip joints at the Middlesex Hospital in 1938 by using a metallic cup fixed to the pelvis using screws, and a hemispherical femoral head fixed to the femur with a bolt. The components were ground to fit accurately together. All the patients suffered from Still's disease and a limited amount of success was achieved. However the prosthesis ultimately disintegrated.

In 1948 Austin Moore in South Carolina introduced a femoral head replacement with a stem that was introduced into the femoral medullary cavity and hence overcame some

of the mechanical problems of the previous designs. Moore had fenestrations in the shaft of his prosthesis to allow bony penetration of the stem and secure fixation.

There were two main problems with the Austin Moore prosthesis. The first was the resorption of the stump of the neck of the femur. Frederick Thompson had noted that after femoral neck fracture the absorption seemed to continue down to the intertrochanteric line and then ceased. He therefore modified the design of the Moore's prosthesis by providing a neck which reached almost to the intertrochanteric line of the femur. He also used a curved stem to accommodate the lateral bow of the femur and used a solid stem because he feared breakage. Because of the need to resect more neck to use his prosthesis he stressed the preservation of the abductor muscles and advocated an anterior approach to the hip joint.

The second major problem with the Austin Moore prosthesis was erosion of the acetabulum with supero-medial migration of the prosthesis into the pelvis. This problem was shared by Thompson's prosthesis.

The problem of the acetabulum was tackled by McKee in Norwich in 1951 who, after his initial designs had failed, used a Thompson prosthesis and articulated it with a cup screwed into the roof of the acetabulum. He found that success depended upon the components remaining tight in the bone. The acetabular component was redesigned in 1961 to allow it to be fixed with acrylic cement.

Acrylic cement, was introduced into orthopaedic surgery after extensive investigations by Wiltse during the period 1952 - 57. It has allowed considerable advances to take place in the development of hip arthroplasty.

Prominent in the introduction of this material to clinical practice and central in the successful development of total hip replacement surgery was the late Sir John Charnley. His appreciation of the mechanics of the hip joint and the properties of materials made him realise that friction between the two metal bearing surfaces of the McKee arthroplasty was a potential cause of failure. He developed the concept of the low friction arthroplasty using a small diameter steel femoral head, articulating with a material of low coefficient of friction (Fig. 1.1). His early choice of polytetrafluoroethylene (P.T.F.E.) proved disastrous because of excessive wear and severe tissue reactions but the later introduction of ultra high molecular weight polyethylene was successful and established the standard material for acetabular cups. Both components were fixed to the bone by a grouting of acrylic cement. There have been design changes in the Charnley prosthesis since its introduction, largely directed at increasing the strength of its stem and preventing loosening. Many other prostheses have been developed but nearly all embody the basic principles illustrated by Charnley.

In the past ten years there has been a reawakening of interest in the idea of interposition arthroplasty at



Fig. 1.1 Femoral and acetabular components
of the Charnley hip arthroplasty.

the hip to avoid the necessity of broaching the femoral medullary cavity (Amstutz, 1977, Wagner, 1978, Freeman, 1978). The approach has followed developments at the knee joint and involves resurfacing both sides of the joint using a metallic cup on the stump of the femoral neck and an ultra high molecular weight polyethylene cup in the acetabulum (Fig. 1.2). Early results were encouraging (Amstutz, 1977) and there was a tendency to use the procedure in young patients to leave the femoral medullary cavity inviolate and suitable for later revisional surgery. However, recent results (Amstutz, 1982, Freeman, 1982) have proved the procedure to be disappointing, the major problem appearing to be with fixation although femoral neck fracture and dislocation have also figured large. Wagner (personal communication) alone continues to report good results but this is probably a reflection of the extreme care he takes in patient selection and his insistence on obtaining as near a normal anatomy as possible, by osteotomy if necessary, prior to the joint replacement. Research is proceeding in this field particularly with regard to cementless fixation (Freeman - personal communication). A major problem that must remain is the difficulty of revising the acetabular side of the joint in the event of failure, since it is always necessary to use a large acetabular cup thus reducing the amount of pelvic bone stock.

The current status of hip arthroplasty is that of a symptomatically successful procedure in a variety of joint diseases. The need for objective assessment of the effect

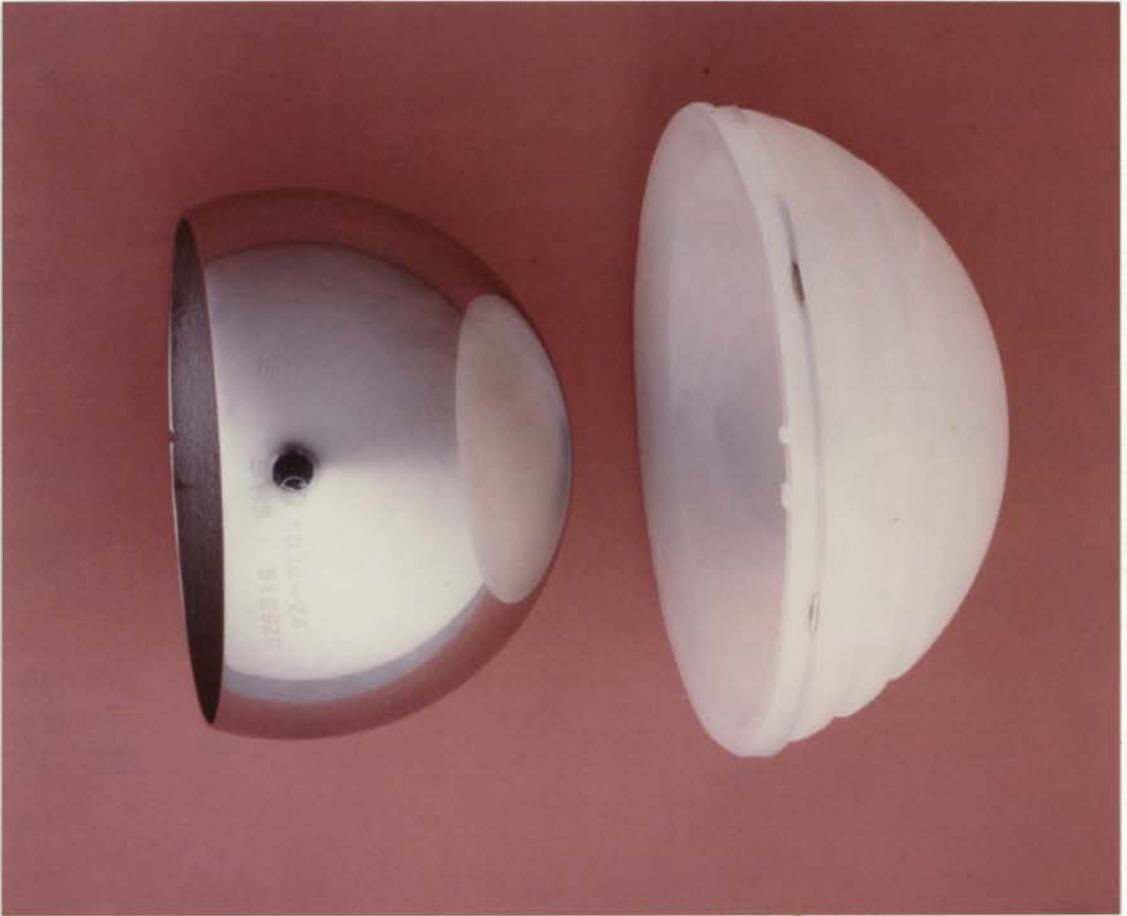


Fig. 1.2 Femoral and acetabular components
of the 'Tharries' hip arthroplasty.

of the joint disease and joint replacement upon function is evident. Locomotion represents the principal functional expression of the lower limb joints and it is therefore relevant to consider the methods used in its study.

1.III LOCOMOTION STUDIES

1. History

Human locomotion has been studied for many centuries and yet we still know remarkably little about how man propels himself about the globe and how pathological conditions affect his progress. In the third century BC Aristotle recognised the importance of muscles with regard to locomotion, linking their contractility with the angular displacement of the body segment to which they were attached. He recognised that this angular displacement of the body segment would be transformed to translational motion, thus allowing the progression of the body. No further advances in the understanding of locomotion occurred until the Renaissance when Leonardo da Vinci's concept of the body as a machine was taken up by Giovanni Alfonso Borelli (1608 - 1679) who applied physical laws allowing him to calculate certain muscle forces and the centre of gravity of the body. He recognised that during gait the centre of gravity moves forwards beyond its supporting base and that the falling of the body is prevented by forward movement of the legs. His important work was presented in the posthumous work, "De motu animalium" (1680).

Further developments had to wait the evolution of the clock into a device suitable for the measurement of time in intervals of seconds; the Kymograph which allowed the recording of mechanical events and allowed them to be timed in intervals of a second using a uniform rotation of the drum and a device such as a tuning fork; and

finally the camera and photographic plates which evolved from their primitive forms in the early 19th century to instruments capable of recording the displacement of moving objects by the 1870's.

The Weber Brothers, Edward (1806 - 1871) and Wilhelm (1804 - 1891) used this new technology to study human gait, their motive being "curiosity and the joy of collaborative effort". They made a great contribution by identifying the phases of gait and their variations with different activities, and by describing the movement of the body in terms of displacement of body segments, a practice which has been used ever since. Unfortunately, their data, based on visual observation alone, was not accurate and it was due to the development of the camera that Professor Etienne Jules Marey (1873) at the College of France in Paris was able to show several errors in the Weber Brothers work using time lapse single exposure photography (Chronophotography) (Fig. 1.3) He thus demonstrated the obvious advantages over visual observations.

At this time Eadweard Muybridge (a photographer hired by Leland Stanford, a Californian industrialist and governor) was engaged in photographing the movements of racehorses using a series of cameras and this classic work led the development of motion pictures. Muybridge also developed the technique of placing a bank of portable cameras at right angles or oblique angles to the subject to obtain three-dimensional recordings. However, although he later used his techniques to record human

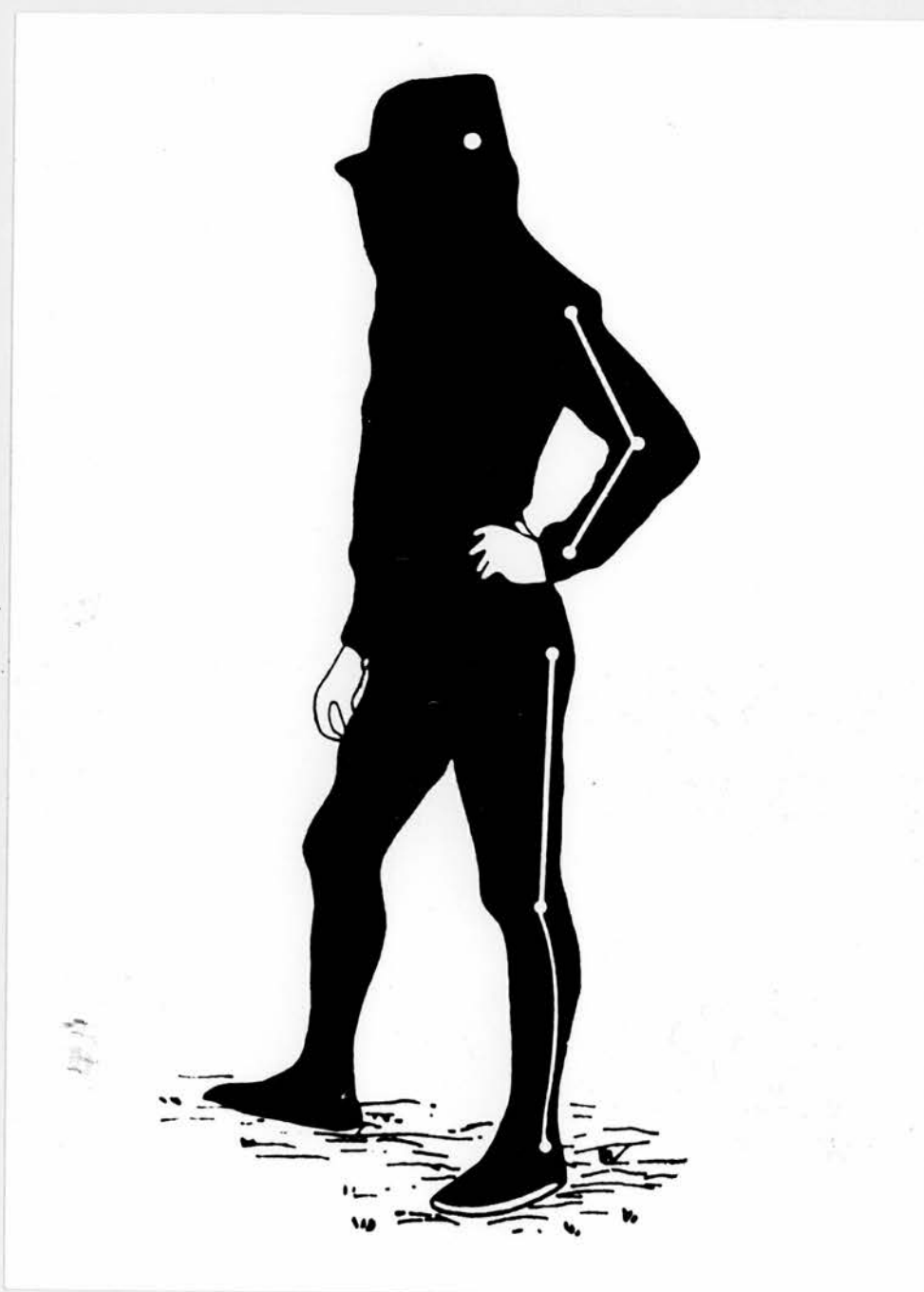


Fig. 1.3a Clothing worn by Marey's
gait analysis subjects.

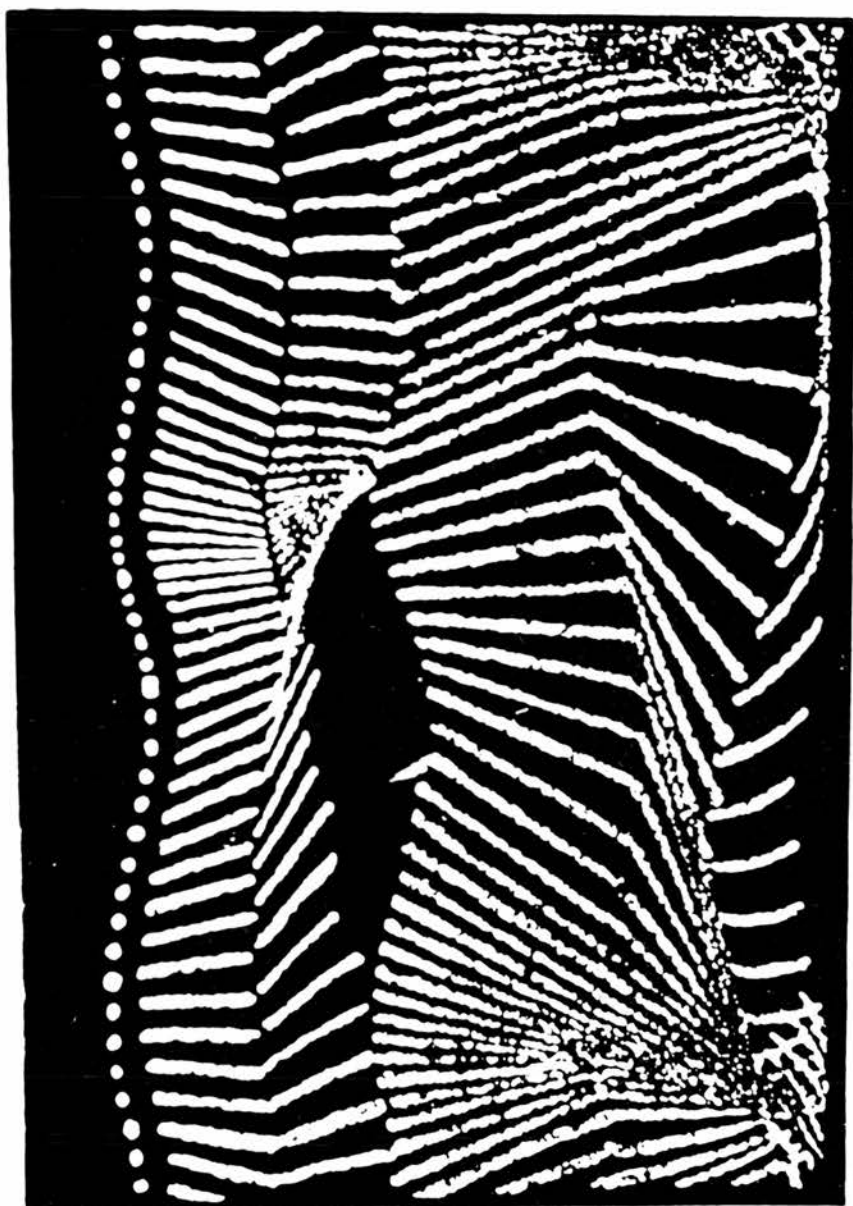


Fig. 1.3b Chronophotography of subject
walking (Marey 1873).

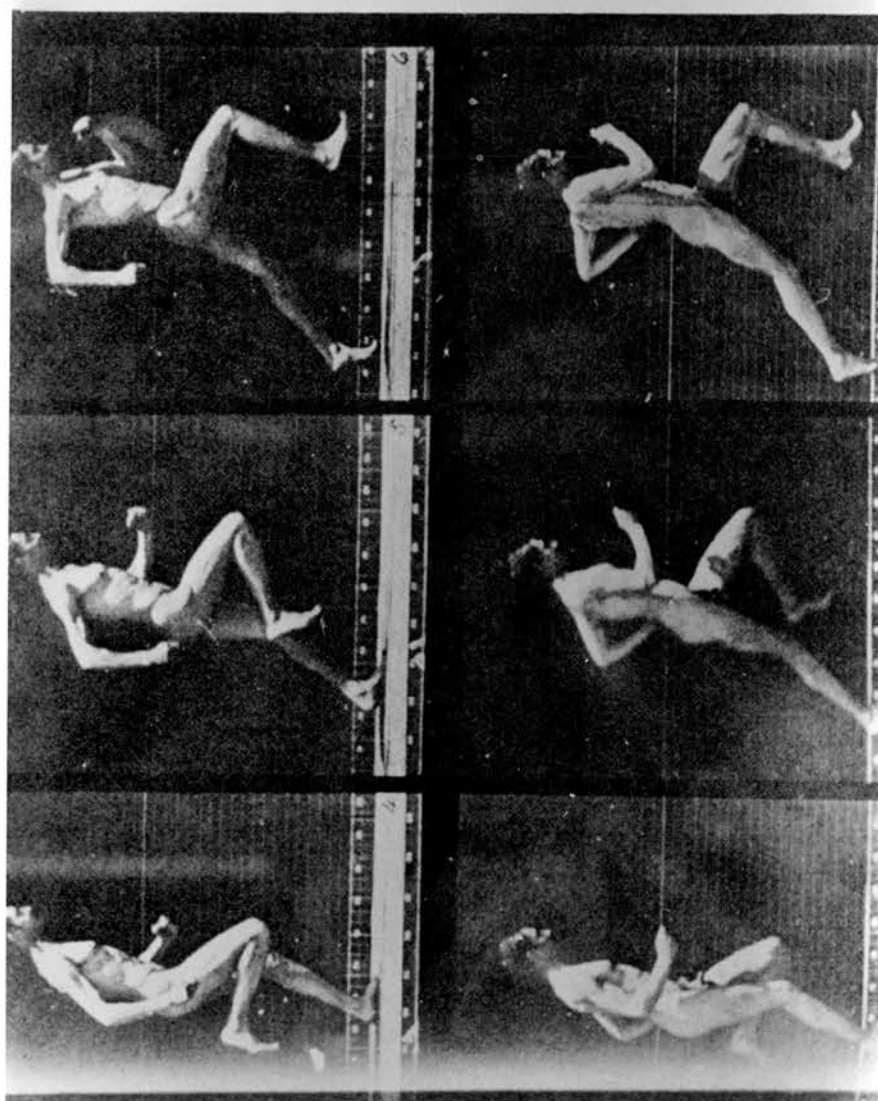


Fig. 1.4 Muybridge's Photographs
of Human Movement
(Muybridge 1979).

locomotion (Fig. 1.4) Muybridge's contribution was more artistic than scientific and no mechanical analysis derived from these studies.

After movements, the next parameter of gait to be measured was the floor reaction forces. Carlet, a pupil of Marey in Paris, published an account of his work in 1872 describing how small bellows were attached to the soles of the subjects' feet and then connected to a kymograph via rubber tubing (Fig.1.5). This allowed both the measurement of the foot-floor force and the events of the gait cycle.

By the end of the 19th century these investigators had provided the language, concepts and techniques to quantitatively record observable characteristics of locomotion. There has been little conceptual advance in this area since then despite considerable technical refinements. However, this century has seen the advances in the concepts of non-observable quantities such as force and energy.

In this area the work of Wilhelm Braune (an anatomist) and Otto Fischer (a mathematician) is of prime importance. In "Der Gang des Menschen", published in six volumes between 1895 and 1901 they presented the results of studies of the energy expenditure and optimisation of locomotion in a foot soldier of the Prussian Army carrying a full pack. Their main contribution was the modelling of the human frame as a dynamic multi-segment mechanism and the calculation of intersegmental forces and moments using the equations describing rigid body dynamics.



Fig. 1.5 Carlet's bellow shoes
and kymograph (Carlet 1872).

They recorded the displacement (movement) data of gait, using intermittently flashing Geisler tubes strapped to various body segments, and made time-lapse photographic exposures of the subjects during gait from four angles. Velocity and acceleration was calculated from differentiation of the displacement curves. The inertial properties of the body segments were measured from three cadaver specimens after making certain assumptions about their configuration. From this data they were able to calculate the forces between foot and floor and the intersegmental resultant forces and moments during the phase of single support. They did not attempt to calculate the forces carried by individual anatomical structures, the problem being then, as it is now, that there were more unknowns (muscles, ligaments and joint forces) than equilibrium equations, making the problem indeterminate (vide infra). The conceptual approach of Braune and Fischer continues to be used in locomotion studies, but technical advances have changed the experimental methods.

The first of these advances was in the development of devices to measure the foot to ground reaction forces. Jules Amar in Paris in 1916 described his "Trottoir Dynamographique" (Fig. 1.6) which was a mechanical device measuring vertical medio-lateral and fore and aft forces by the deflection calibrated springs under the platform.

Elftman, an anatomist described his mechanical forceplate in 1938. It comprised two platforms, the lower platform being supported by springs to measure the

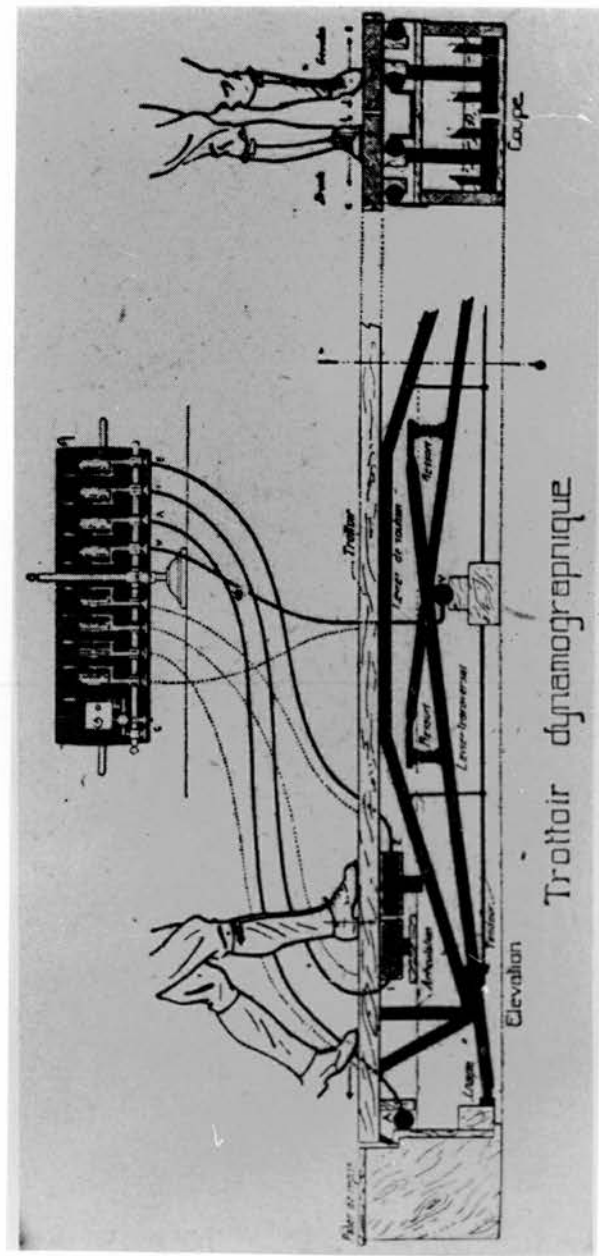


Fig. 1.6 Trottoire Dynamographique (Amar 1916)

vertical forces and the upper one separated from the lower by ball bearings, and restrained by springs to measure the horizontal components of force. Displacements of the platform were measured by a lever system and recorded by a high speed cine camera.

The problem with both of these force measuring devices was that they moved when the subject walked over them causing him to alter his gait. This problem was overcome by a group from the College of Engineering and the Medical School of the Berkeley Campus of the University of California as a part of their major contribution to the study of locomotion. Cunningham and Brown described the force plate in 1952. It consisted of a top plate mounted on four strain gauged columns. Loads applied to the top plate caused no discernible movement but produced strains in the columns which resulted in signals from the gauges which were proportional to the applied load. The most commonly used force plate today (manufactured by Kistler Instruments A.G. of Switzerland) differs little in concept from that of Cunningham and Brown, but uses three pairs of piezo-electric discs in each column, each pair being sensitive to load in one of the three principal axes.

The work of the Californian group is of major importance. With the motive of improving artificial limbs for war veterans they set out to study normal locomotion and to apply their findings to these subjects. They developed a system for the study of gait comprising interrupted light and cine film studies, glass walkways,

EMG's and force plate studies. They collected information concerning the displacements and rotations of the limbs in space, velocities and accelerations, the external forces on the limbs and the phasing of muscle action during gait (University of California, 1947, Eberhart and Inman, 1951). Their use of metal pins drilled into skeletal locations to act as markers is of particular note.

Bresler and Frankel (1950) reported upon the three-dimensional analysis of the forces and moments in the leg during level walking in four subjects using the force plate of Cunningham and Brown. Skin markers were used to define the joint centres and cine photographs were taken along the axis of progression and at right angles to the subject to obtain three dimensional coordinates. They used the fundamental principle of all force plate studies, established by the Californian group, that the measurement of the forces and moments referred to the surface on which a subject is walking provides the most convenient way of completely describing the forces acting on the limb during locomotion. By resolution of the resultant forces and moments to an equivalent load system, related to the force plate, the forces acting on any part of the body can be computed from a knowledge of the orientation of that part with respect to the centre of the force plate and the magnitude of the loads on the plate. In this way they calculated the intersegmental forces and moments acting about the ankles, hips and knees of their subjects. They compared the computation of external moments with and without inertia

components and found that the effect of gravitational and inertial forces on the fore and aft moments at the ankle, knee and hip were small throughout the stance phase of the gait cycle. However, Andrews (1976) and Tooth (1976) have expressed the opinion that to estimate internal forces and moments at the hip (i.e. muscle and joint forces) inertial effects cannot be ignored. One other observation of relevance from the work of Bresler and Frankel is that although the floor reaction forces represented a valid first approximation to those at the ankle and knee considerable errors could be introduced at the hip by such an approximation.

Much of the fundamental work in the calculation of joint loads has been performed in the Bioengineering Unit at the University of Strathclyde by Paul (1965, 1966, 1967a, 1967b, 1968, 1970, 1971, 1972), Morrison (1967), and Poulson (1973). The method used in all of these studies involved the recording of kinematic data using cine-cameras, one in the axis of progression and the other perpendicular to this axis, and the acquisition of kinetic information by the use of a Cunningham and Brown force plate. Three dimensional analysis was performed to obtain the external joint moments at the hip or knee. Paul calculated the muscle forces required to balance the external moments at the hip using a simple muscle model and thereby calculated the load across the hip joint (vide infra) (Fig. 1.7).

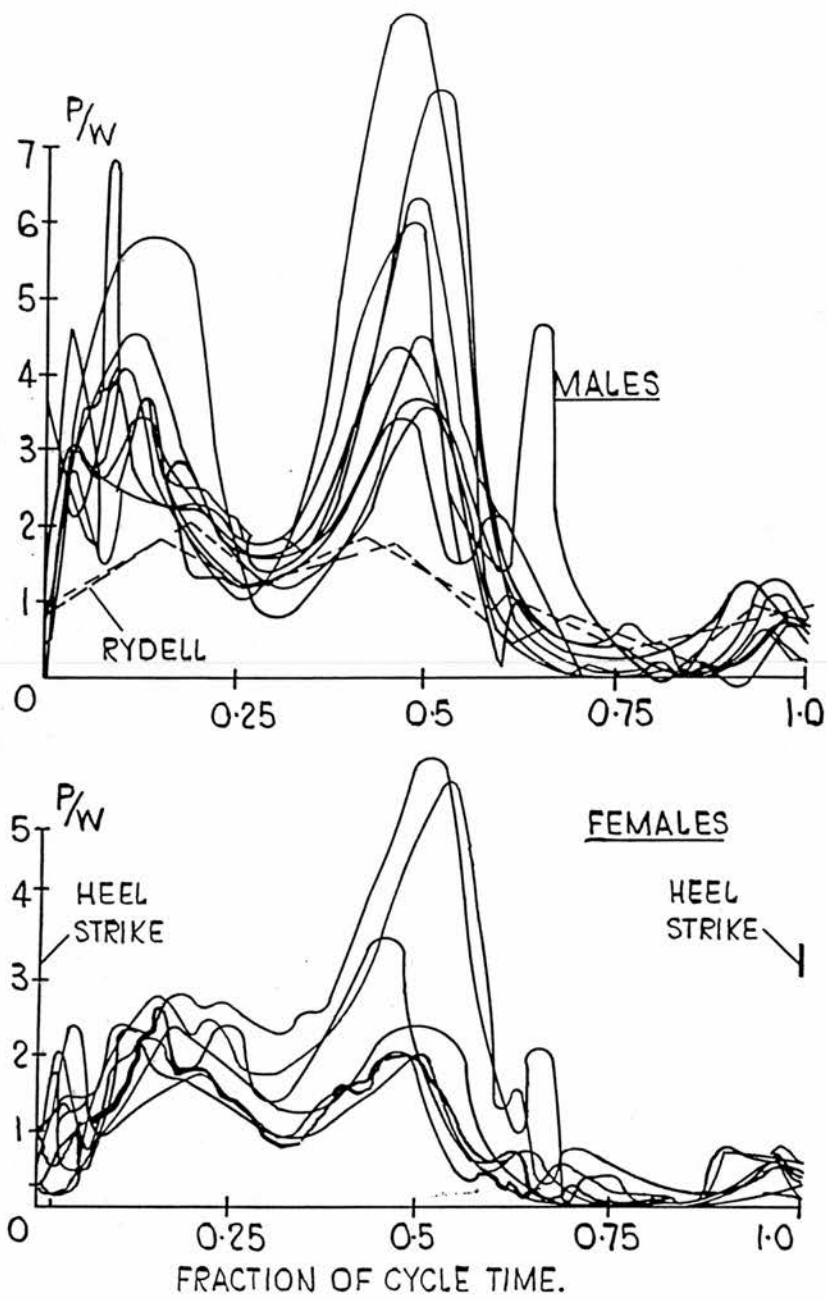


Fig. 1.7 Paul's hip joint force curves
for normal subjects (Paul 1967)

Morrison performed the same type of analysis to calculate the loads at the knee joint (Fig. 1.8) and Poulson combined both of these approaches in the investigation of muscle activity and forces at hip and knee joints in the same limb for various activities in normal subjects.

2. Instrumented Implants

The validity of the indirect methods of joint load assessment alluded to above can be evaluated by comparing the results obtained for hip joint force by Paul with those obtained by the use of implanted instrumented prostheses.

In his now classical work, Rydell (1965, 1966) implanted an Austin Moore like femoral head replacement into two patients. The prostheses had strain gauges bonded into their necks (Fig. 1.9). Wires were led to a subcutaneous site at the time of operation and were brought to the surface for connection to a measuring system at a second operation six months later once normal activities had been resumed. Rydell obtained peak values for hip joint force during walking of 2.8 times body weight which compare with Paul's values of 4 to 6 times body weight when walking speed and stride length are taken into consideration (vide infra).

English and Kilvington (1979) have reported on the use of an instrumented femoral component of a total hip replacement designed by English (1975). The prosthesis resembles the standard Charnley femoral component but has

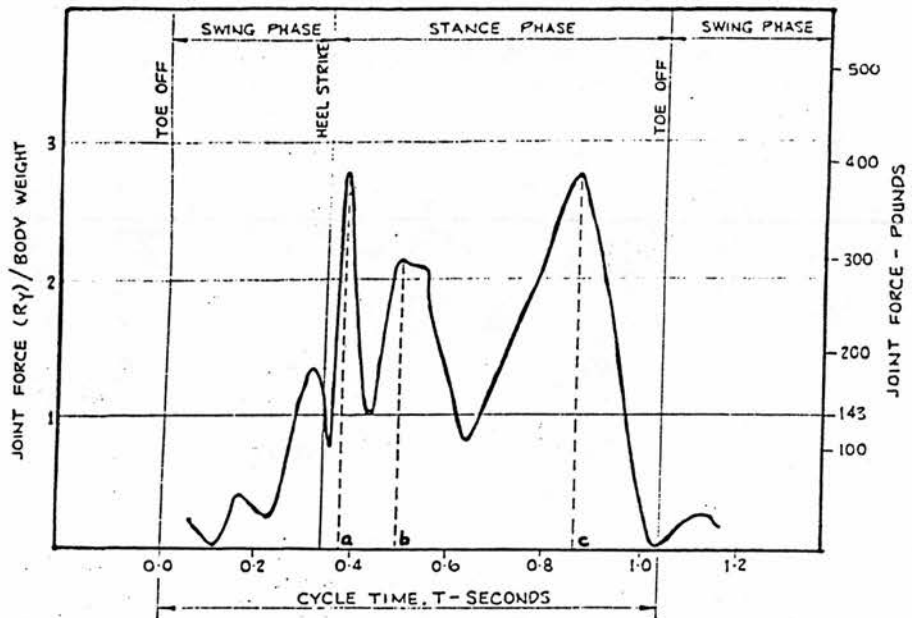


Fig. 1.8 Morrison's curve of variation in knee joint load during stance phase (Morrison 1967)

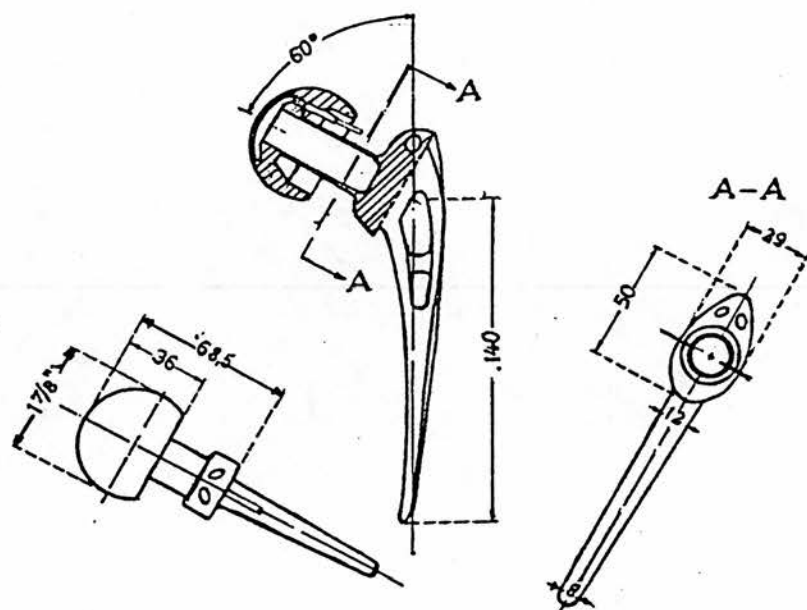


Fig. 1.9 Rydell's instrumented hip prosthesis (Rydell 1965).

a collar, a modified stem design and a smaller offset (Fig. 1.10). The strain gauges were mounted in the neck of the prosthesis on a "piston-in-cylinder" arrangement, and transmission of output was by the use of a telemetry system incorporating the miniature battery powered F.M. transmitter activated by the taping of a loop antenna to the skin over the hip. The device was used in one patient and recordings were made over a forty day period post-operatively (Kilvington and Goodman, 1981) at which stage the signal output failed. The results for stance phase hip loading at 2.0 times body weight and 2.54 times body weight for the one legged stance compares well with Rydell's findings although it must be pointed out that Rydell's prosthesis was designed to measure bending forces whereas the English prosthesis was designed to detect the mean axial force acting down the neck of the prosthesis. Additionally, the English prosthesis was articulating with a high molecular weight polyethylene cup and was inserted with lateral trochanteric displacement (English, 1975b).

These results represent the load across an altered joint or joint replacement and, particularly in the case of the total joint replacement, the condition for which the prosthesis was inserted may have resulted in secondary changes in ligaments and muscles which will have precluded normality.

A study reported by Brown, Bernstein and Frankel (1982) using instrumented fixed angle nail plates in the

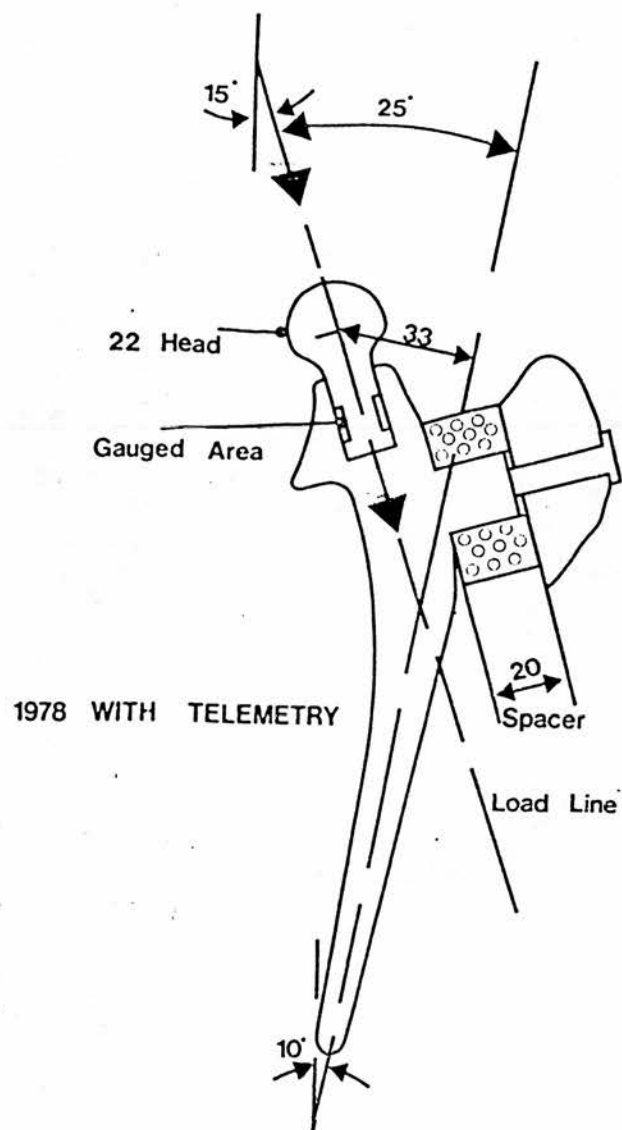


Fig. 1.10 English's instrumented femoral component (English and Kilvington 1979).

management of intertrochanteric fractures has the potential to provide more meaningful information concerning joint force, although this quantity has not yet been calculated. However, such studies are expensive, invasive and complicated by serious ethical considerations. Indirect methods of assessment provide a much more practical method of assessing joint force and much work is in progress to improve their accuracy (Crowninshield & Brand, 1981).

3. Other Approaches to the Study of Gait

Methods of recording gait parameters have been described by Contini, Gage and Drillis (1965). Mention must be made in passing to the use of interrupted light photography (Murray, Drought and Kory, 1964), stereo-photogrammetry (Bullock and Harley, 1972), ultrasound (Finlay, Wuta and Hall, 1973), goniometers (Johnston and Smidt, 1969, Lamoreux, 1971) polarised light goniometers (Grieve, 1969, Mitchelson, 1974), visible and infra-red light emitting diodes (Soderberg and Gabel, 1978, Andriacchi et al, 1979) and the use of accelerometers (Cavagna et al, 1963, 1966, Gage, 1964).

4. The Need for Computers

The major problem with most systems has been the time consuming nature of the data reduction. Bresler and Frankel studied four subjects and reported that the time for data reduction in each case was 250 to 500 man hours! Winter et al (1972, 1974) developed a television

gait analysis system to tackle this problem. It was a two dimensional system and data from videotape was fed via an interface into a computer for analysis. The utilisation of computers has done much to advance the study of locomotion and they are central to the application of experimental methods to the clinical situation.

Jarret (1976) and Andrews (1972) have developed a three dimensional television-computer gait analysis system at the Bioengineering Department at the University of Strathclyde by the design of an interface which correlates television movement data with force plate data in digital form and passes them on to the computer. This is the system which has been used in the work reported in this thesis (Chapter 3).

1.IV. MUSCLE FUNCTION DURING GAIT

Considerable advances have taken place in the understanding of muscle physiology and muscle activity during gait since the time of Braune and Fischer and this work will now be summarised.

1. Muscle Mechanics

The muscle length - tension relationship under galvanic tetanic stimulation (Ralston et al, 1953) is a well-known and accepted physiological principle. In 1910 Fick introduced the concept that a muscle's physiological cross sectional area is linearly related to the muscle's maximum force exertion capability. Other investigators have confirmed this relationship and quoted constants of proportionality varying between 0.4 and 1.0 MN.m^{-2} (Morris, 1948, Ikai and Fukunaga, 1968). However, the physiological cross sectional area of a muscle has not been precisely defined, particularly in relation to shortening or lengthening, or to the muscle's structure. It is now generally accepted as being the muscle's volume divided by its length (average cross sectional area). Very little information appears in the literature with regard to this parameter (Gross, Chao and An, 1976, Rippinger, Chao and Stauffer, 1980) and it must be noted that the values obtained for a muscle's physiological cross section depend upon the method of measurement having particular regard to the position of the skeleton and its effect on muscle length.

The muscle fibre composition (e.g. slow twitch and fast twitch fibres) of a muscle differs not only from muscle

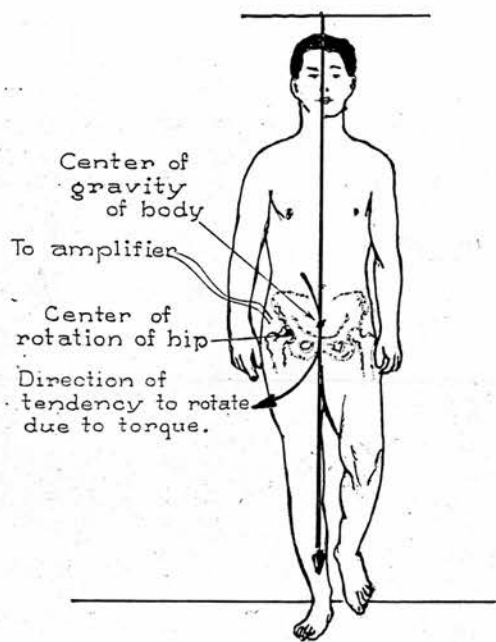
to muscle within an individual but also within the same muscle in different individuals and it appears that the percentage of fibre types is related to strength (Dons et al, 1979) and endurance (Thorstensson and Karlsson, 1976). These differences may account for the variations in the reported proportionality constants relating muscle force to cross sectional area.

2. Electromyography

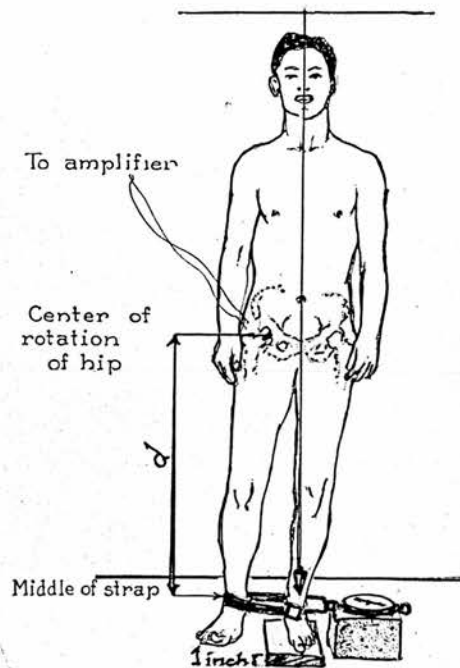
In his 18th century study of electrical current Galvani first reported the relationship between electrical activity and muscular activity. Duchenne (1864) stimulated individual human muscles with galvanic currents and determined their function.

The first attempts to assess muscle function during gait were those of Richard Scherb in Zurich. In 1927 he reported on the assessment of these muscles using manual palpation but later used E.M.G.

The early finding of a relationship between E.M.G. activity and muscle force (Dern, Levene and Blair, 1947) gave some hope that muscle forces could be quantified by external measurements. However, this work concerned EMG in subjects performing maximum voluntary movements and the quantitative relationship between EMG and muscle force in muscles acting non-isometrically has proved to be complex. Moreover, even with isometric contraction the relationship is not simple. Morrison (in Paul, 1972) studied the effect of isometric extension of the knee against resistance and found that at low loads there was increasing activity



Torque = Weight of body x distance from center of rotation of hip to sagittal plane of body.



Torque = $d \times \text{scale reading}$

Fig. 1.11 Inman's method for measuring hip abductor muscle torque (Inman 1947).

of rectus femoris with no activity of the vasti, but at a definite load activity commenced in the vasti and increased thereafter in both the vasti and rectus femoris.

Inman (1947) studied the abductor muscles of the hip in a static study. He analysed the forces acting in the one-legged stance - a two dimensional analysis. The integrated action potential was recorded from the tensor fascia lata, the gluteus medius and gluteus minimus using surface and needle EMG electrodes. The subject then transferred his weight to the other limb, under which was placed a small block to allow the contralateral foot to clear the floor as it was abducted against a spring balance (Fig. 1.11). The moment arm was measured on the patient and the action potentials were recorded for a series of abduction forces as measured by the spring balance. The integrated action potential was plotted against abductor moment and the value of the latter, corresponding to the action potential in one-legged stance, was obtained. These values are erroneous. Even if it is accepted that with maximal isometric contractions muscle force is proportional to EMG activity, Inman's experiment involves neither maximal force (except on the extreme occasion) or isometric conditions since he took no precautions to control muscle length.

Work is proceeding (Hof and Van Den Berg, 1981, a,b,c and d) attempting to correlate muscle force with EMG but at present ignorance of the complex relationship of EMG to muscle force during activities such as walking precludes any use of such a correlation.

Many workers have used EMG to investigate the phasic activity of muscles during gait (Joseph and Battye, 1966, University of California, 1947) (see Chapter 3). Milner, Quanbury and Basmajian (1970) have emphasised the effect of walking speed upon the phasing of muscular activity. Knowledge of the phasing of muscular activity is an essential prerequisite for the construction of a model for the behaviour of muscles.

3. Muscle Modelling

At any joint there is considerable functional redundancy with regard to muscular action. Twenty-two muscles cross the hip joint, clearly illustrating this fact. Joint function during locomotion occurs as a result of the temporal phasing of the muscles acting about it, and because of the redundancy of muscles the body can exercise considerable discretion in the selection of muscle groups to perform a particular activity.

In order to determine the forces in muscles and ligaments the external intersegmental forces and moments must be distributed amongst them. Because there are more muscles (unknowns) than equilibrium equations (three each for intersegmental forces and moments) the situation is statically indeterminate and to solve the distribution problem, a mathematical model of muscle function has to be constructed. There are three basic approaches to this problem of modelling.

i. Functional grouping and reduction method

With this approach the redundant muscle force

variables are either reduced or combined, usually on a functional basis, to permit only six unknown structures to transmit force at any instant. Solutions to the equilibrium equations can therefore be obtained. This is the basis of the method used in this study although use is also made of the other approaches to be described, and it will be discussed in detail in Chapter 3.

ii. Systemic Elimination

If subsets of the multiple variables are selected at random and unique solutions to the equilibrium equations are obtained, multiple solutions will be generated (Chao, Opgrande and Axmear, 1976). Many of these are then eliminated on the basis of set criteria, i.e. a) joint contact forces become negative b) muscle force is compressive c) a muscle force exceeds the estimated strength according to its physiological potential (vide supra) and d) the antagonistic muscle carries abnormally high force. All permissible solutions are combined to produce the range of muscle and joint forces for activity. The proponents of this technique hold the view that the physiological solution of the distribution problem falls within the region bounded by all possible admissible determinate solutions.

iii. Optimisation Method

In this method some process relevant to joint or muscle function is minimised or maximised subject to a set of constraint conditions in an attempt to obtain a determinate solution to the problem, without significant

simplification of the functional anatomy. The wide choice of muscles for a given activity has been mentioned and it seems intuitively reasonable to assume that control of human locomotion involves optimisation of body function. For example, with a painful hip joint, if pain is due to joint surface pressure, the appropriate optimisation criterion would be the minimisation of joint force. The choice of criterion to solve a particular problem may not be obvious. Because of the extremely repetitive nature of the calculations necessary for optimisation techniques their use has only been practical since the introduction of high speed computers.

Many reports of optimisation procedures appear in the literature, almost all of the early communications using linear criteria because of the ease of mathematical calculation. The work of Seireg and Arvikar (1973, 1975) is of relevance to lower limb function during walking. They investigated the effect of minimising a) the sum of the force magnitudes, b) the moment caused by the ligaments, c) the vertical component of articular surface force, and d) muscle work. They concluded that the preferred optimisation function was a minimisation of the sum of muscle forces plus four times the moment carried by the ligaments. They gave no reason for using four times the ligament moment. They also made considerable simplifying assumptions viz. only the vertical component of the floor reaction force was considered, and its configuration was taken to be a plateau rather than the

two peak and trough pattern common to nearly all reports of floor reaction forces in normal gait. Their values for the hip, knee and ankle joint forces were 5.4, 7.1, and 5.2 times body weight for the hip, knee and ankle joints respectively. Their high value for the knee joint load when compared to that of other investigators (Morrison, 1967, Poulson, 1972) may well be a reflection of their assumptions.

Non-linear optimisation was first applied to the muscle forces during gait by Pedotti, Krishnan and Stark (1978). They compared the use of linear and non-linear functions using several criteria. They indicated a preference for the objective function of the sum of the muscle stresses (muscle force divided by the physiological cross-sectional area) squared, in view of the correlation of the predicted muscle forces with reported EMG activity (vide supra).

Crowninshield and Brand (1981) have reported on the use of the inversely non-linear relationship of muscle force and the possible contraction duration as a criterion for the optimisation of muscle forces. They suggest that muscle selection so as to maximise endurance is physiologically reasonable during many normal activities. They stress that their criterion is not applicable to all forms of locomotion and cite activities occurring to minimise pain. They emphasise the need to use sound physiological constraints in the evaluation of the optimal criterion, e.g. the correlation of muscle force predictions with EMG activity cannot be taken as proof of the validity of

the criterion unless the muscle force predictions also correlate with the muscle size.

However, all of these optimisation approaches to the solution of the distribution problem suffer from the fact that no single optimal criterion has been validated physiologically and no single criterion can be expected to fulfil the need in different joints or activities.

All of these approaches to the solution of the indeterminate problem are subject to limitations. The most crucial of these as highlighted by Patriarco et al (1981), is the accuracy of the calculation of the external joint moments which depends upon precision in the determination of joint angles, and ground forces.

Secondly, the need to assume no antagonistic muscle activity, since the rationale for the selection of such activity by the neuro-muscular system is unknown, may result in erroneous conclusions with regard to muscle forces.

Thirdly, the commonly used technique of validating the muscle forces by EMG is open to criticism in view of the work of Eberhardt et al (1947) which demonstrated considerable variation in inter-individual EMG activity, and that of Brand (Personal Communication, 1982) indicating considerable cycle to cycle variation in a given subject.

The ultimate source of limitations in present gait models is the incomplete nature of our knowledge of the physiological function and role of individual muscles during the gait cycle. Resultant joint loads are calculated from muscle forces by vector addition.

The variability in the values of muscle forces cited in the literature results from the varying techniques of measurement and muscle models used. However, as Crowninshield and Brand (1981) have indicated, these various studies result in joint contact force predictions which are far less variable than muscle force predictions (Fig. 1.12). Higher values for joint force will result from those methods that predict antagonistic activity.

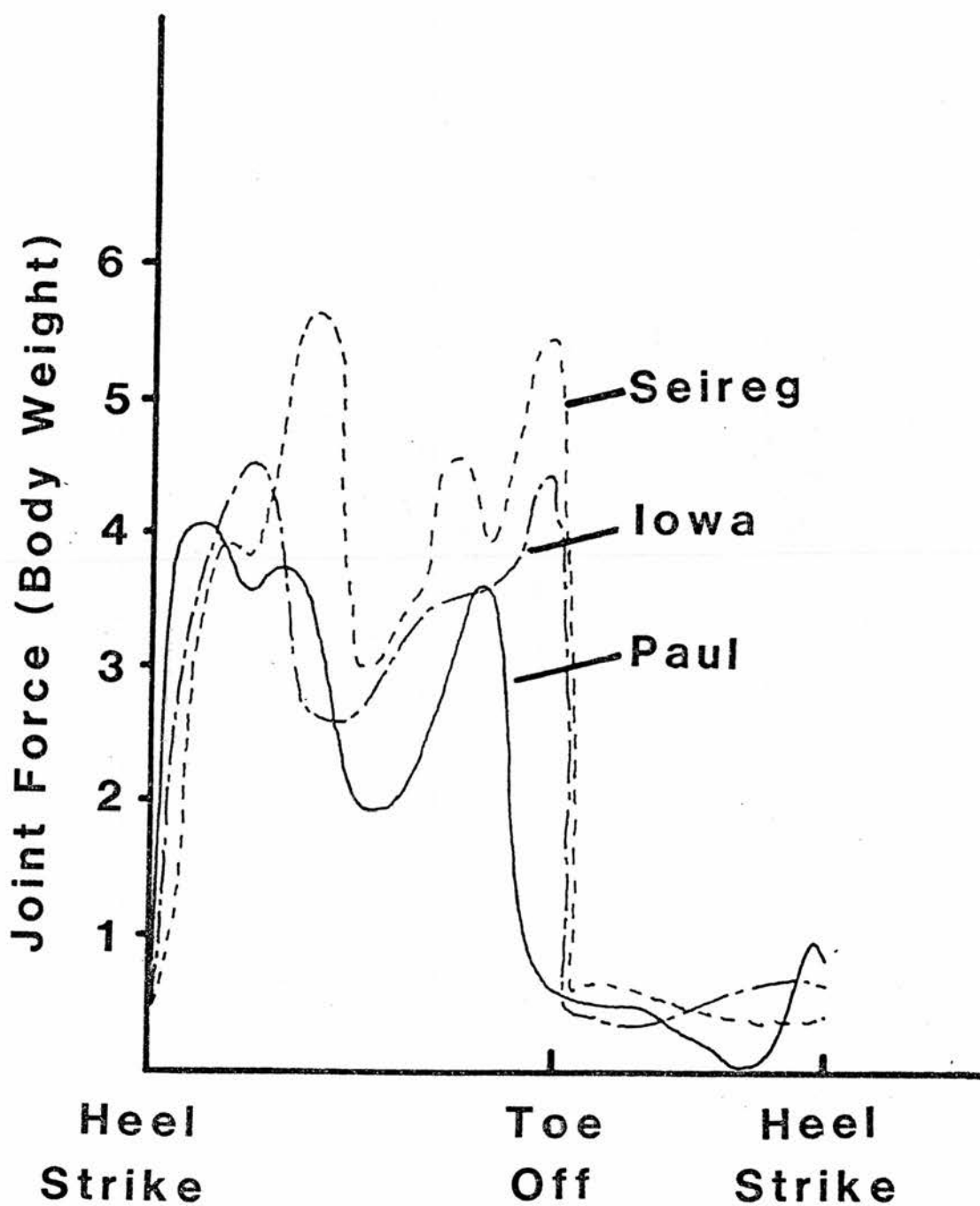


Fig. 1.12 Magnitude and Pattern of Hip Joint Force during Level Walking determined by the Use of Different Muscle Models. (Crowninshield and Brand 1981).

1.V JOINT LOAD AND JOINT DEGENERATION

The joint load is of particular interest in relation to the aetiology of degenerative joint disease. Freeman (1972) suggested that the initiator of degeneration was a chemical change in the matrix of the articular cartilage which interfered with its ability to carry load. A review of the weight-bearing properties of cartilage led him to the conclusion that the compressive stiffness of articular cartilage derived from its mucopolysaccharide content whilst its collagen content was principally responsible for its tensile properties. Abnormalities in both of these components preceded fibrillation.

Fibrillation may occur as a result of multiple fatigue fractures in the surface collagen. This follows an imbalance between the mucopolysaccharide content of the cartilage together with the mechanical properties of its collagen and the magnitude and frequency of compressive stresses applied to the surface. This might be expected to result in the rupture of the surface collagen.

The loss of articular cartilage creates a local increase in the stress to which the subchondral bone trabeculae are subjected and Freeman (1972) has estimated the stress in some trabeculae to be about 24MN m^{-2} using an estimate of 3100N for the joint load, a figure based upon Paul's work. Swanson, Freeman and Day (1971) have shown that at least 60MN m^{-2} applied for 1,000,000 cycles was needed to produce failure in trabecular bone. Therefore, if Freeman's estimate is correct, fatigue

fracture is unlikely to occur in the normal femoral head. However, if local stresses were abnormally high and present for a sufficient length of time fatigue failure might be seen.

More recently, Chrisman and his colleagues at Yale University (1982) presented what they claimed to be the first evidence clarifying the link between mechanical injury and the early biochemical changes in the osteoarthritic process. They used a computerised pendulum device to deliver a moderate, precisely measured, reproducible impact to the surface of canine femoral head cartilage, designed to simulate the forces involved in dashboard impact in a road traffic accident, or a severe blow to the knee in an athletic event. The lipids of the cartilage were analysed by gas chromatography and a four-fold increase in arachidonic acid was found in the phospholipid pool compared with controls. Arachidonic acid had been shown, in previous studies by Chrisman, to be a precursor of prostaglandins and a major component of the double-layered phospholipid cell membrane. Its central role in Chrisman's proposed "osteoarthritic cascade" (Fig. 1.13), the final common pathway of cartilage degeneration indicates the significance of the findings.

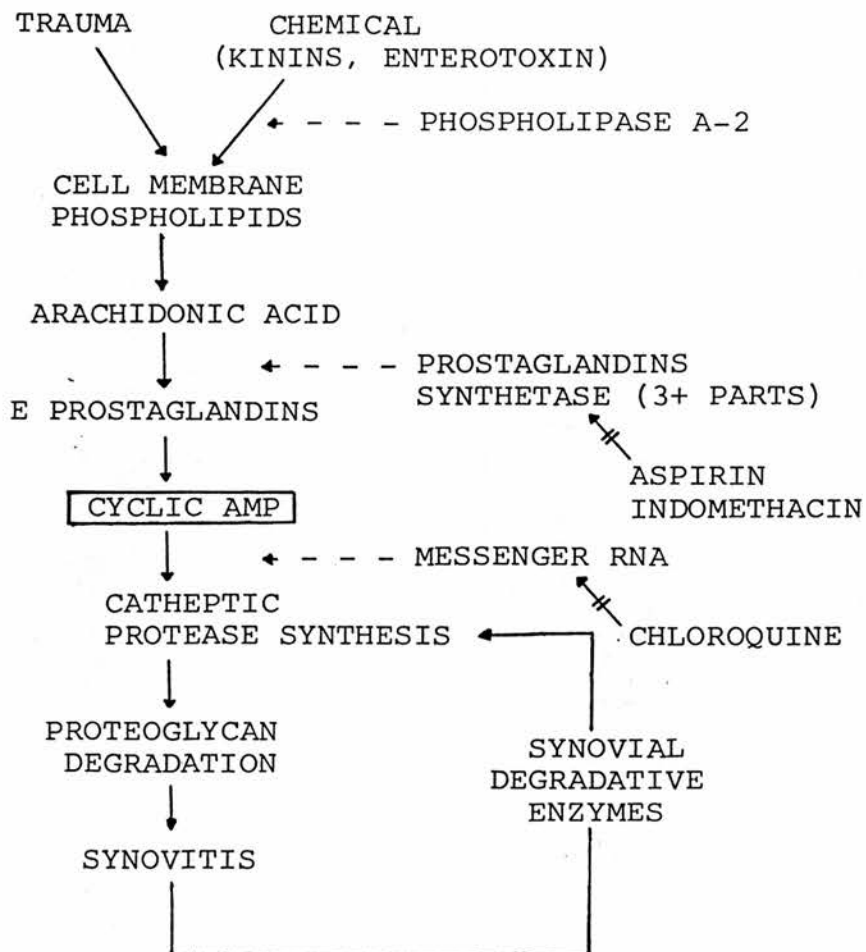


Fig. 1.13 The Osteoarthritis Cascade
(Chrisman et al 1981)

1.VI STUDIES OF GAIT IN DEGENERATIVE HIP DISEASE AND ARTHROPLASTY OF THE HIP

1. Clinical Methods

Surgeons have looked for methods of assessing the functional status of their patients occasioned by hip disease and its change after arthroplasty, for many years. Many clinical methods have been devised for this purpose (Merle, D'Aubigne and Postel, 1954, Lazansky, 1967, Harris, 1969). All involve the calculation of a functional score and place greatest emphasis upon pain, whilst also considering such gait factors as walking distance, the use of walking aids and the ability to climb stairs. Whilst having a place these assessments suffer from the limitations that they are subject to large observer differences, particularly with regard to the assessment of pain, and they are dependent upon the patient's history. This, especially in the case of walking distance, can be wildly inaccurate (McGavin et al, 1978). Finally the ultimate functional score does not reflect in which sphere improvement has or has not taken place.

Walking is an indicator of the function of the lower limb and as we have seen many methods have been developed for its study. The application of these methods to the problem of hip joint disease and arthroplasty will now be discussed.

2. Energy Requirement Studies

The estimation of oxygen requirements to perform simple exercises such as level walking for a fixed distance, and climbing stairs has been used as a measure

of function. Early studies suggested that there was no difference in oxygen consumption before and after hip arthroplasty but McBeath et al (1980) and McNicol et al (1980) demonstrated that the oxygen consumption required to cover any given distance, standardised for the weight of the patient, changed significantly after arthroplasty. This improvement was greatest during the first six months after surgery and was most marked in those patients who were most disabled pre-operatively. A direct correlation between the oxygen consumption and the walking speed was demonstrated and it was suggested that "the self-selected velocity" of a patient was a satisfactory indicator of the walking efficiency of that patient. However, a review of their work resulted in further communication (McNicol et al, 1982) explaining their findings purely on the basis of the patient's walking speed, there being no relationship to efficiency of gait. This approach would therefore appear to be extremely limited in application and productive of very little information regarding the cause of any failure to attain function or the loss of function.

3. Biomechanical Studies

The human musculoskeletal system can be adequately described as a complex arrangement of mechanical linkages. The motion of the limb segments is achieved by coordinated muscle contractions together with smooth and congruent articulation of the joints. Total joint arthroplasty is used in an attempt to restore the integrity of a defective system through mechanical and surgical repair of the diseased components.

The mechanics of the hip joint depend upon its anatomy, the state of its articulating surfaces, and the function of the muscles acting about it.

About 80% of the load across a normal hip joint is a consequence of muscle action (Paul, 1967). The activity of the muscles depends upon their anatomical arrangement which is determined by the skeletal anatomy (1.V.1). In osteoarthritis of the hip and after total hip arthroplasty the skeletal anatomy is altered and muscle function is affected. Additionally, in osteoarthritis of the hip, pain may cause inhibition of certain muscle actions. Therefore it can be predicted that there will be an alteration in the limb movements and the hip joint load in both diseased and replaced hip joints. Hence a mechanical approach to the assessment of gait in its role as an indicator of hip joint function would be logical.

a) Time and Distance Parameters of Gait

The time and distance parameters of gait - walking speed, cadence, stride length, stance and swing phase durations are the most simple to measure and have been extensively investigated for the normal subject (Murray, 1967). They represent an obvious and easily recognisable indicator of function.

Several authors (Murray, 1966, Grieve, 1968, Paul, 1970, Andriacchi, 1977) have reported upon the way in which the walking speed influences limb movements, muscle activity and joint forces and the change in the relationship of the

phases of the gait cycle that it brings about. Since most patients with joint disease and many with hip arthroplasty walk more slowly than the "normals" that have been studied it is important to take the differences in walking speed into consideration when interpreting the results of gait studies.

Murray and her colleagues (1971) have attempted to characterise the limp of coxalgia and have reported a slower walking speed than in normal subjects (94 cm.s^{-1} versus 195 cm.s^{-1} for a normal group) with shorter steps and slower cadence, and, most notably, a greater inequality of successive step lengths. This results in a slow and irregular progression (see Chapter 2) comprising longer steps with the painful limb than the sound limb and a shorter duration of the stance phase of the gait cycle on the painful side.

Smidt and Wadsworth (1973) confirmed this pattern, but also noted that the velocity and cadence of a group of normal individuals walking slowly was similar to that of a group of patients consisting of several with bilateral hip disease, and almost half who used two crutches for walking. Closer examination of their results reveals a higher cadence and velocity than the slow normals for those patients with hip disease who used no walking aid, or one cane or crutch, illustrating the multifactorial nature of the problem under study and the fallacies that may result from the comparison of the means of homogeneous and heterogeneous groups.

Similar findings are reported by Stauffer (1974), and in a recent study by Walls (1981) working in Northwick Park using a conductive mat linked to a microcomputer. The patient wore conductive tape on the sole of his shoes and walked across the mat, time-distance information being calculated automatically by the computer.

Because these parameters are so easily obtained many studies of patients undergoing total hip replacement have been published. Murray and her colleagues in Milwaukee (1975) studying patients treated with McKee-Farrar hip arthroplasty before, six, and twenty-four months after surgery, found that there was an improvement in the walking speed, cadence and stride length post-operatively. There was also more time spent in stance phase on the involved limb than pre-operatively producing greater symmetry of gait. The "velocity index", the rate of the average velocity during the period of single limb support on the untreated limb over that on the limb which was operated upon, was used as an indicator of the uniformity of forward progression. All but one of the patients with unilateral disability showed more uniform progression at two years after hip arthroplasty and all but nine of the patients with bilateral hip disease showed more uniformity.

Stauffer and his colleagues in Iowa (1974) studied a group of 25 patients with hip disease, 15 of whom also had symptomatic abnormalities in other joints, prior to and six months after Charnley hip arthroplasty. They found an increase in walking velocity after the arthroplasty

(mean 35.5 cm.s^{-1} to 61.8 cm.s^{-1}). This was achieved by a small increase in cadence and a large increase in stride length. The increase in the time-distance parameters after hip arthroplasty approached but did not reach normal, even taking into account the slight effects of age in decreasing cadence and stride length (Murray, 1969, Crowninshield et al, 1978). Interestingly, in contrast to Murray, they noted an increase in the percentage of the gait cycle occupied by the stance phase on the side of the involved limb pre-operatively which did not change post-operatively. They did not give any data regarding the sound limb, although considering the composition of their patient group fewer than 50% of their subjects would have had sound limbs. Their results cannot, therefore, be taken as an indicator of hip joint function. However, the authors did attempt to determine the effect of multiple joint disease by coding the degree of involvement numerically. There was an inverse correlation between the degree of multiple joint involvement and pre- and post-operative velocity, cadence and stride length.

Although the time-distance parameters are sensitive indicators of gait alterations, by themselves they do not indicate the cause of the gait abnormality. For this, measurements of ranges of limb movement (Kinematic studies) and ground reaction forces (Kinetic studies) are required.

b) Kinematic Studies

Murray and her group in Milwaukee have been responsible for most of the kinematic data which is available concerning gait in hip disease and after hip arthroplasty.

Their studies have been concerned with limb and trunk movements during gait and have included information about isometric muscle strengths of hip musculature and some kinetic data derived from instrumented walking aids.

The system of analysis used was interrupted light photography with an overhead view through a mirror. The patient was required to walk in semi-darkness on a raised platform illuminated by a strobe-light flashing at 20Hz. and it is questionable whether what was recorded represented a true record of the patients' natural gait.

In degenerative hip disease with a painful hip joint, a study of 25 men with unilateral hip pain and no other bone or joint abnormality (Murray et al, 1971) showed that these patients used a smaller flexion-extension range at the involved hip joint than at the sound hip or hips of normal men. This reduced range was almost entirely the result of limitation of hip extension in the later part of stance, but it did not correlate with the magnitude of any hip flexion contracture that was present and was probably a pain avoidance mechanism. At the sound hip the range of flexion-extension was similar to normal but this hip showed more flexion and less extension than the hips of normal subjects during the entire walking cycle.

Other body movements were also of interest, all but one patient showing an excessive lateral shift of some part of the body towards the weight-bearing side during stance on the painful limb. All patients showed descent of the pelvis on the sound side during stance phase on the painful limb irrespective of the degree of lateral shift, and even if the

Trendelenberg test was negative. This has been interpreted as the result of a lengthening contraction of the abductor muscles. During stance phase on the sound side there was abnormal elevation of the pelvis on the painful side, and it was suggested that this might represent an attempt to improve the clearance of the foot on the painful side during its swing phase.

At the knee joints, the contralateral knee moved in a similar fashion to the normal, whilst the knee on the painful side remained in extension until the latter half of stance when it gradually flexed.

Stauffer, Smidt and Wadsworth (1974) describe similar abnormalities of hip movements, although in less detail, in patients with hip and other lower limb joint pathology (vide supra) using an electro-goniometer strapped to the limb of the patient. They also comment that there was no significant correlation between the clinically measured, passive hip motion and the motion used in gait. The data of Murray, Gore and Clarkson (1971 - vide supra) is in agreement with this. This latter study failed to determine whether the limp of coxalgia was due to pain or loss of mobility but as Stauffer comments, the fact that in his study the correlation between the clinically measured range of movement and motion used during gait is more positive post-operatively (vide infra) suggests that pain must play a significant part in the causation of the observed movement abnormalities. At this time pain is not a quantifiable parameter, although Nagai and his co-workers in Kyoto (1976) report that the integration of

the walking traces of the forces acting on the foot during several steps (measured using shoes fitted with 16 component force transducers) produces an almost linear trace, the gradient of which they feel corresponds to the level of walking pain. The lesser the gradient the greater the degree of pain. This approach remains to be evaluated and at present we must accept that pain is a variable which may make the interpretation of gait analysis data in patients with painful joint conditions difficult to the extent of being valueless in some cases.

After hip joint replacement considerable changes take place in the kinematics of gait. Most of the information in this area again comes from the Milwaukee group. In their analysis of one hundred patients with hip joint disease treated with McKee-Farrar hip arthroplasties, some with bilateral involvement and some with other lower limb joint disease (Murray et al, 1975) they noted significant changes in the ranges of joint motion used during walking over a two year follow-up period. The range of flexion-extension at the hip joints increased post-operatively in most cases although the patients tended to use more hip flexion and less extension than normal subjects. The lateral lurch of some part of the body towards the involved limb during its supportive phase persisted post-operatively although to a lesser degree than pre-operatively. Further studies by the same group (Gore et al, 1977, Murray et al, 1979, 1981a, 1982) have confirmed these findings in patients treated with Muller and Charnley arthroplasties (vide infra).

Stauffer, Smidt and Wadsworth (1974) found an increase in the range of flexion extension used during gait by their patients after arthroplasty (mean 12.1 degrees to 32.8 degrees post-operatively). This was still less than the range of hip movement reported by Murray (1964) for normal men, but greater than that reported by Murray for the McKee-Farrar arthroplasty. There was a statistically insignificant increase in the range of abduction-adduction used during walking (4 degrees to 5.8 degrees) leaving this much less than the range reported for normal subjects (11 degrees). Similarly, there was only a very small gain in the range of rotation used at the hip post-operatively (6.6 degrees to 7.4 degrees) and the question was raised, but not answered, as to whether this lack of recovery of movement was a result of the operative technique. As already mentioned this study is difficult to interpret with regard to the function of the hip arthroplasty because of the multiple joint problems of many of the patients. Additionally their system of measurement - a goniometer using potentiometers and strapped to the patient - was liable to inaccuracy owing to movement between the patient and the device. This might be expected to have a significant effect upon the accuracy of the rotatory movement measurements in particular. Finally neither the design of the arthroplasty nor the geometry resulting from its insertion was considered as an explanation for the lack of recovery of movement.

The angle - angle diagram approach to the description of the kinematic features of gait (Grieve, 1968) in which movements of one joint are plotted against those of an adjacent joint, was applied to patients before and after total hip replacement by Milner (1973, 1974) (Fig. 1.14). The subjects dressed in black catsuits with reflective tape marking the limb segments and were photographed using stroboscopic flash photography as they walked. In order not to obscure the markers the subject's arm had to be held to his side. Additionally a tachometer was used to dictate a constant walking rate. This affects the validity of the data but the method is useful and gives an indication of deviations from "normality" or perhaps, more correctly, the asymmetry of the gait. However, it provides little more information than this.

c) Kinetic Studies

Information about limb movement patterns draws attention to abnormalities and the observer is left to speculate about the site of the abnormality, and the way in which the joint is functioning to produce the observed abnormal movements. The function of a joint depends upon muscle action as well as mobility and the best indicator we can obtain of muscle action during gait, without invasive techniques, is the measurement of the ground reaction forces, subsequently relating them to the limb movements and dimensions to calculate intersegmental moments.

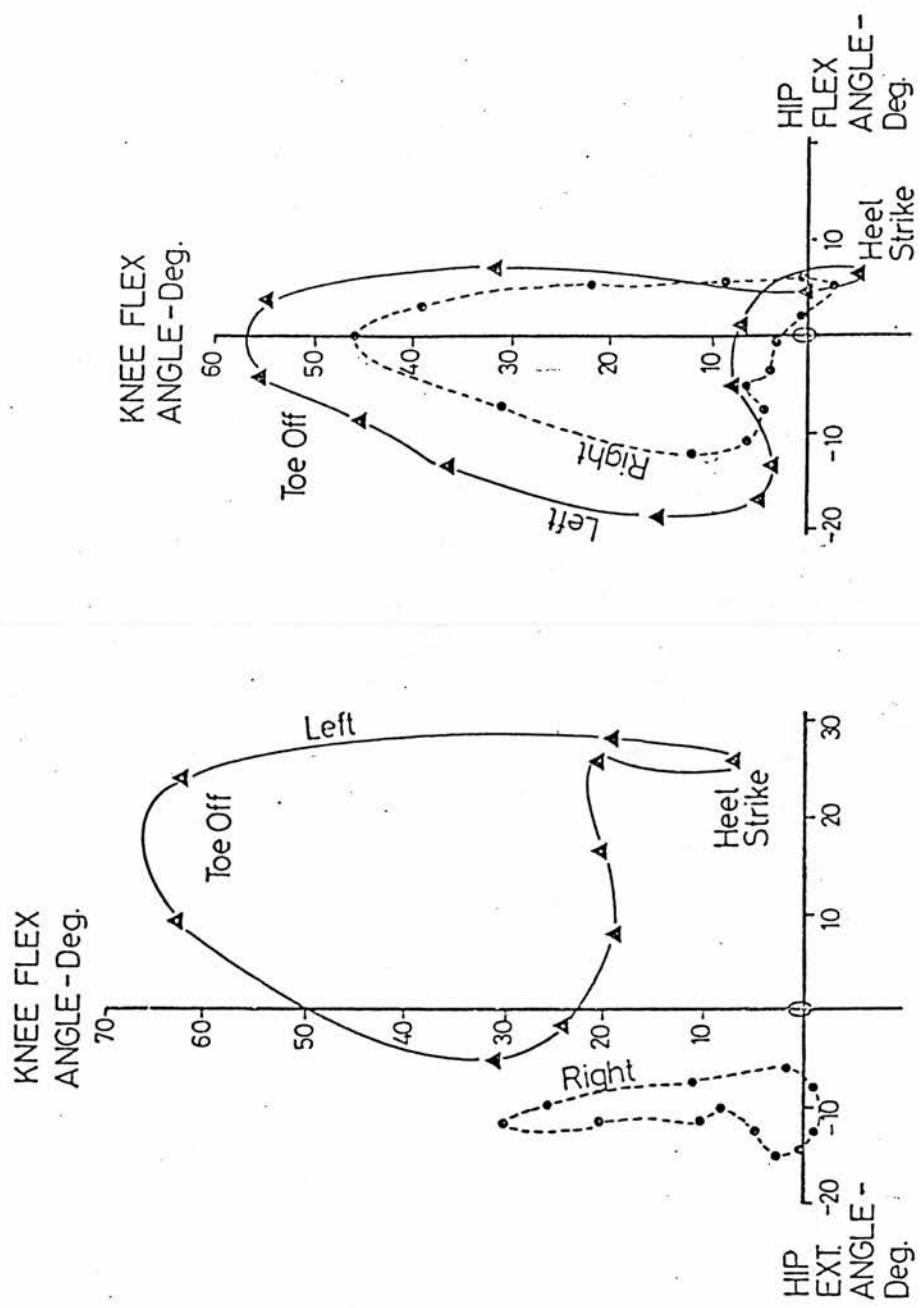
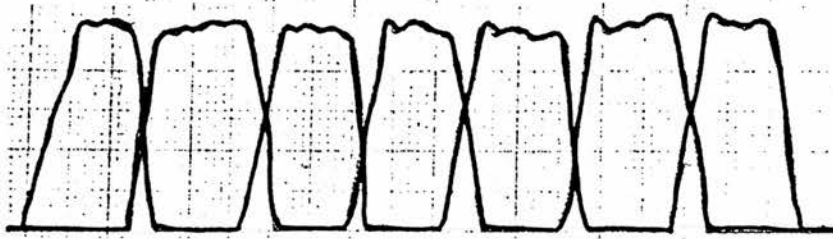


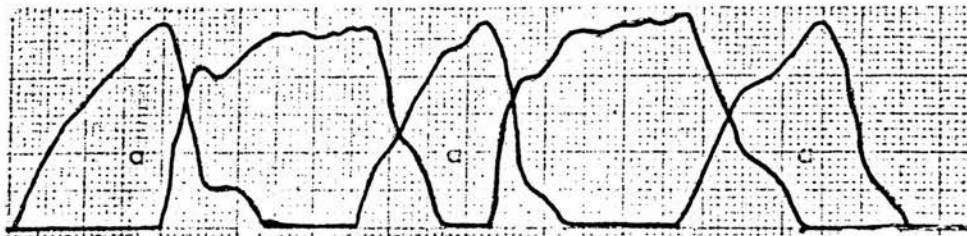
Fig. 1.14 Angle-Angle Diagrams
 (a) before and (b) after total hip replacement
 ----- affected limb ——— healthy limb
 (Milner 1973).

i) Ground Reaction Forces

The background to the development of the measurement of ground reaction forces has already been described. The first description of the measurement of these forces in patients with hip disease and hip arthroplasty can be attributed to Charnley and Pusso in 1968. Using an instrument designed by Skorecki in Manchester in 1961 (Skorecki, 1966) they studied the vertical reaction force at floor level during gait. The device enabled the forces from several sequential steps to be analysed, and used a mechanical system. Several normal subjects were studied and features of the normal trace were established. These were :- equal duration of double support phases, regular occurrence of all pulses, and symmetry of the force pattern. Two types of abnormal gait were identified; symmetrical abnormal gaits and asymmetrical gaits. An example of a symmetrical abnormal gait is that seen in patients with severe bilateral lower limb arthritis. All asymmetrical gaits are abnormal and are well exemplified by that of unilateral hip disease where the duration of the stance phase on the sound limb is prolonged. In this situation the shape of the pulse differed on the sound side there being loss of the usual two peak curve and replacement by a plateau (Fig. 1.15). The authors make the point that nothing of the characteristic shape of the pulse on the healthy side remains unless the opposite side is 100% normal. The magnitude of the peaks must always be the same unless a walking aid is being used. The



(a) Normal



(b) Unilateral hip disease
(Charnley and Pusso 1968)

Fig. 1.15 Traces obtained using the
Gait Machine of Skorecki (1966)

results were being used in a semi-quantitative way, rather like an E.C.G. tracing, to aid the clinical decision-making with regard to the assessment of patients for hip surgery, and to assess the progress of a patient's disease or joint replacement.

Jacobs, Skorecki and Charnley (1972) extended the use of the gait machine to pathological gaits and used a computer programme to analyse certain parameters measured from the traces. Fourier Series were also used to aid in the interpretation of the wave forms. A pattern was defined for normal gait but eight wave forms were described for the pathological gaits (all subjects with hip disease) and it was noted that the division between these groupings was not easily identifiable. An attempt was made to correlate the state of the hip joint with the wave form but no useful conclusions were reached.

Unfortunately, this method does not permit the evaluation of forces at the hip joint itself, and disease in the foot, ankle or the knee can influence the results. Also the authors' contention that the most valuable component of the floor reaction force is the vertical component must be questioned, having particular regard to the modes of action of the hip muscles, and their important contribution to hip joint load (Paul, 1967).

Khodadeh, Stokes and Whittle (1981) have extended this approach by using two "Surrey" force plates, which act in a similar manner to Skorecki's instrument, allowing the analysis of the vertical component of the ground

reaction in sequential steps, and two Kistler force plates which permit the evaluation of all components of the ground force. They studied three groups of patients: those with unilateral hip joint disease, those with bilateral hip disease and one hip replacement, and patients with untreated bilateral hip disease. They found a correlation between their quantitative measurements and a clinical assessment. In the unilateral hip disease group there was asymmetry of both the timing and the magnitude of the floor forces, with a slower time to peak loading and a lower magnitude of loading on the affected side. The bilateral group exhibited a symmetrical gait pattern with no significant differences between the limbs. After Charnley hip joint replacement the unilateral group showed a more symmetrical pattern and the symmetry of the bilateral group remained unchanged. Despite having the capability to record the antero-posterior and medio-lateral components of the floor reaction force no mention was made of them in this communication.

Smidt and Wadsworth (1973), however, reported upon all components of the floor reaction force in patients with hip disease. They found the same reduction in magnitude of the vertical force to less than body weight, and reported peak force and aft forces very much lower than those seen in normals (0.5 - 4% body weight versus 2 - 16% body weight in normals). The medial forces were similar for both groups but there was an extremely low lateral force seen in the patients with hip disease. They suggest that patients govern their

walking speed to reduce the rate of loading after heel strike and increase the rate of unloading, in order to minimise the intensity of the pain experienced during the stance phase of gait. If this were established pain might be correlated with the rate of force loading at joint level.

In an extension to this study, Stauffer, Wadsworth and Smidt (1974) reported similar findings and compared these with their findings six months after a Charnley hip arthroplasty. They found an improvement in the magnitude and rate of loading of the vertical component, an increase in the magnitude of both fore and aft forces, a change attributed to post-operative increase in stride length and walking velocity (vide supra), and wide variability in the medio lateral peak forces both pre- and post-operatively.

All of these studies of ground forces are subject to the same criticism as that of Charnley and Pusso, namely that there is no evidence to correlate ground reaction forces with hip joint loading.

A more descriptive and precise parameter can be calculated by combining the measurements of floor reaction forces with the three-dimensional positions of the joints to calculate the resultant moment acting at the joint throughout the gait cycle. The external moments are the net effect of all the external forces acting on the limb segments to produce rotation about any given joint. The laws of mechanics require that the external moments at a joint be balanced by internal moments that are equal

in magnitude and opposite in direction. These internal moments are primarily generated by muscle forces acting at a distance from the joint centre.

The external moments can therefore be taken to be indicative of muscle action and related to joint loading.

(ii) Joint Moment and Load Estimation

Andriacchi and his colleagues in Chicago (1980) have investigated 20 patients who had undergone successful hip arthroplasty at least one year previously. No details were given about the state of the patients' other joints. They assessed the resultant moments about the hip joint using an optoelectronic system for motion analysis (Andriacchi et al, 1979) and a piezo-electric multi-component force plate linked to a mini-computer. The magnitude and pattern of both movements and moments were approximately normal for this group but differences were obtained when the patients were separated on the basis of the length of their abductor muscle lever arm, measured from radiographs. When the lever arm was shorter than the unoperated side by an average of 0.9cm the moments and movements were virtually normal but where the lever arm approximated to that of the other side there was a reduced range of flexion-extension at the hip (27 degrees versus 38 degrees) and the peak flexion moment just after heel strike was "statistically higher" than in the other group (Fig. 1.16). Interestingly, despite separation of the patients on the basis of abductor muscle lever arm no significant difference could be found in the magnitude of the abduction-adduction moments. This study draws attention to the influence of the anatomy upon the parameters of gait,

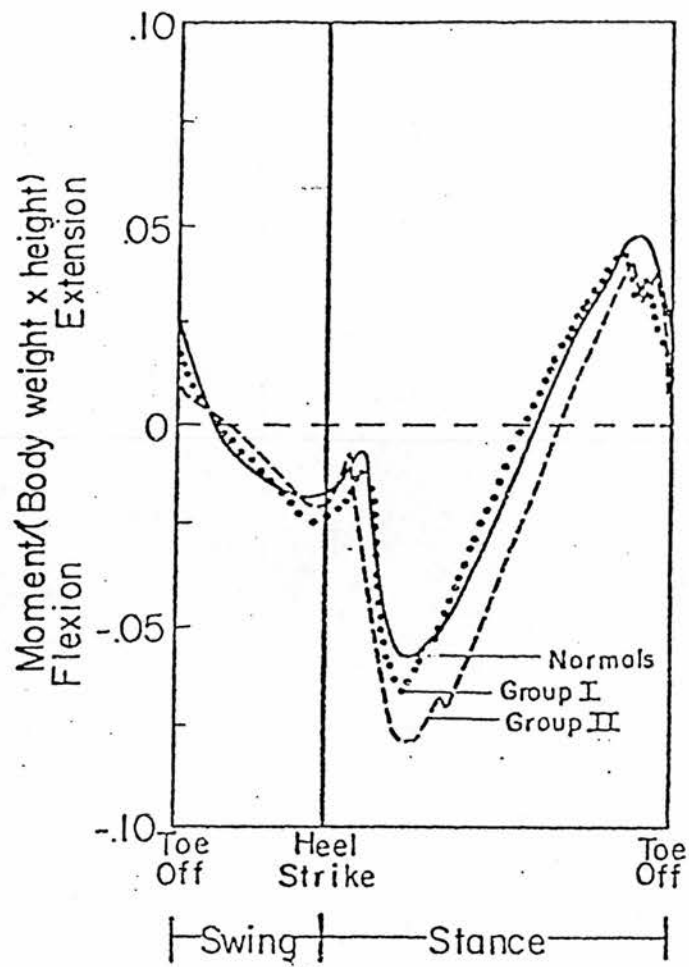


Fig. 1.16 Curves for flexor-extensor moments about the hip joint, after total hip replacement

GROUP I Lever arm on replaced side shorter than unoperated side

GROUP II Lever arm on replaced side longer than unoperated side (Andriacchi et al 1980)

although by considering only one of these in the presence of many interactive anatomical variables the correlation found may have been fortuitous and may not imply a cause and effect relationship. This subject is discussed further in Chapter 1.VII.

The only study of which this author is aware that investigates the joint loads at the osteoarthritic hip joint is that of McGrouther (1975), performed in the Bioengineering Department of the University of Strathclyde using the same system as that described by Paul (1967) and Morrison (1967) for their investigations of the normal hip and knee joints. McGrouther calculated the external moments and joint forces in the hips of 15 patients with unilateral hip disease and no other lower limb joint or lumbar spine abnormality. He found that the magnitude of joint loading was similar to that seen in normal subjects, as calculated by Paul, but the pattern was abnormal (Fig. 1.17). There was no single pattern characteristic of the degenerative hip. Similarly, the magnitude of the forces at the contralateral hips was no different from that at the diseased hip but the pattern was again abnormal, and this was considered to be a result of the gait rather than to indicate early disease in the contralateral hips.

Another communication from McGrouther (1974) describes the presence of abnormal moments and loads about the ipsilateral knee joints of three patients with osteoarthritis of one hip joint (Fig. 1.18). He suggests that this may be due to excessive hamstring activity related to the presence of a hip flexion contracture. He also noted that the centre

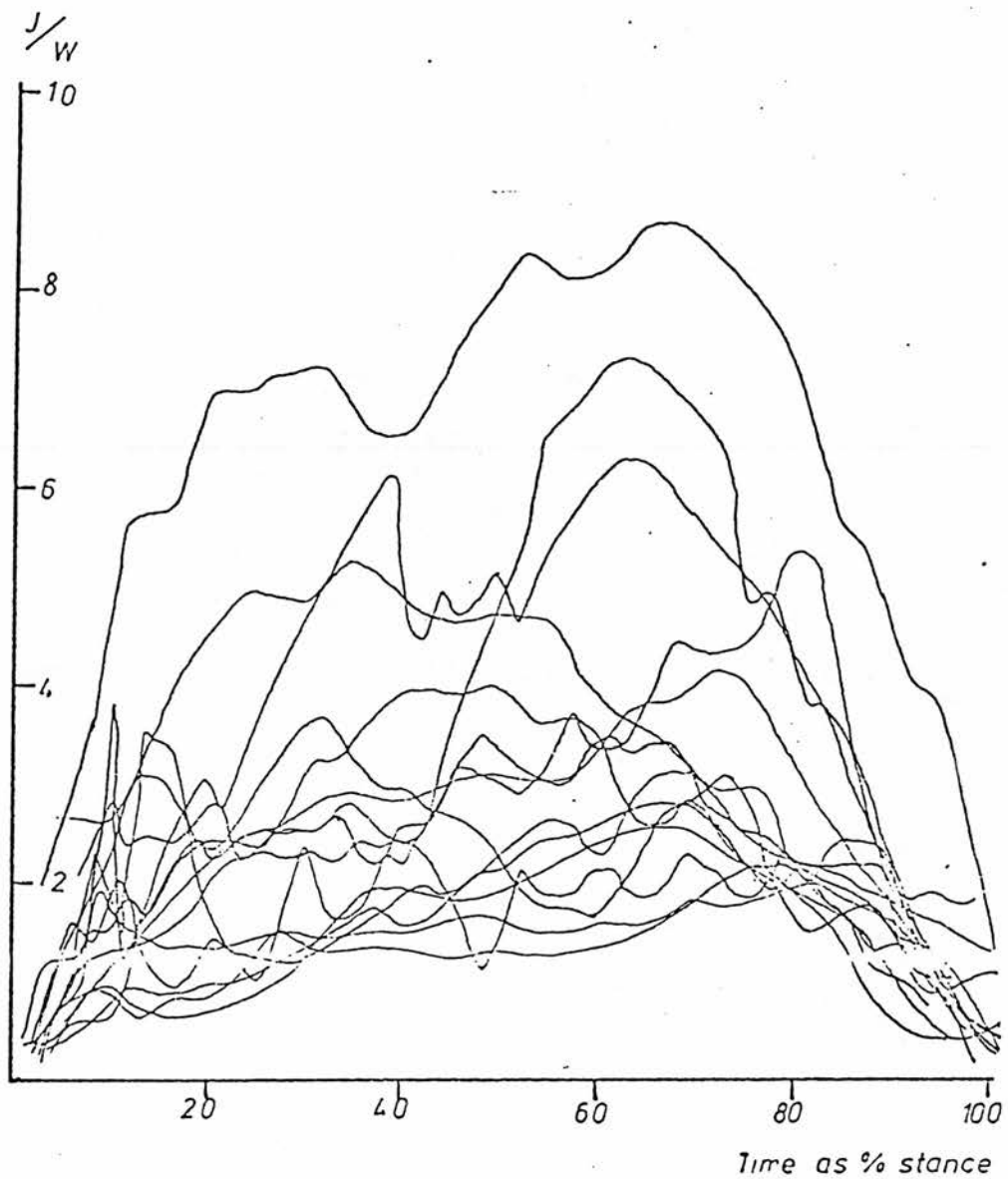


Fig. 1.17 Loads at osteoarthritic hip joints (McGrouther 1975).

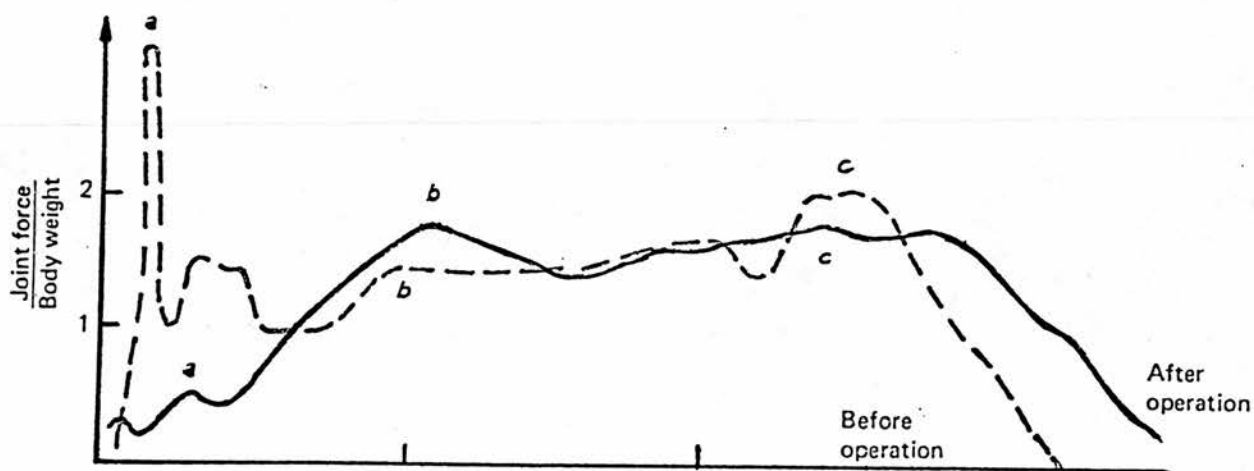


Fig. 1.18 Loads at ipsilateral knee joint of patient with osteoarthritis of the hip joint before and after total hip replacement (McGruther 1974).

of pressure at the knee was in the lateral compartment in contrast to the medial compartment in normal subjects (Morrison, 1967). This he attributed to the antalgic gait pattern in which the patient leans over the affected hip in an attempt to reduce abductor muscle action and hip joint force.

That the antalgic gait did not produce a reduction in peak hip joint force is of interest. In his patients McGrouther noted that the abductor moment was very small or negative (implying activity in the adductor muscles) during the stance phase, although the joint load was not reduced, because of activity in the hip extensors or flexors. This illustrates the caution which must be exercised in applying the theory of one-legged stance forces to the gait cycle (*vide infra*), and challenges the role of the antalgic gait in pain relief.

Although McGrouther mentions that the hip joint centre position was measured from radiographs he made no attempt in his modelling to allow for the change in relationship of the muscles to the hip joint centre. Indeed the hip joint muscles were related to the hip joint centre by three-dimensional proportionality scaling factors. Therefore, although he may have allowed for the anatomical change at the joint due to disease he did not include the effect of this upon the mechanics of the muscles in his calculations.

The observation made with regard to the loads at the knee joint are clearly relevant to the clinical observations of degeneration in knee joints following total hip replace-

ment surgery (Hamblen, 1976) and become even more important when it is noted that after total hip replacement the ipsilateral knee joint force in McGrouther's three patients reverted to normal magnitude and line of action (McGrouther, 1974). Further study of this phenomenon by way of investigating the loading of all lower limb joints in patients with hip disease or arthroplasty is clearly required.

Apart from the studies of Rydell and English (Chapter 1.III 2.) there are very few reports in the literature of loads acting across joint replacements. Again, virtually all the work in this area has been carried out at the University of Strathclyde (McGrouther, 1973, 1974, Paul and McGrouther, 1975, Paul, 1976, Tooth, 1976). The reports of Paul and McGrouther concern the moments and forces at the hip joint after hip arthroplasty in three female patients. The pattern of abductor moments about the affected and contralateral hips were abnormal both before and after surgery, but increased in magnitude post-operatively. Changes in the flexor moments were more variable. Assessment of loads revealed resultant hip joint forces within the envelope of the magnitude of normal values as described by Paul (1967) (Fig. 1.19) but abnormal in pattern, none showing the two peaks and trough. It was suggested that the irregular plateau pattern produced may have been a temporary stage in recovery or may have represented an end point due to permanent loss of muscle function. Again this study deals with the mechanics of the replaced hip joint as if



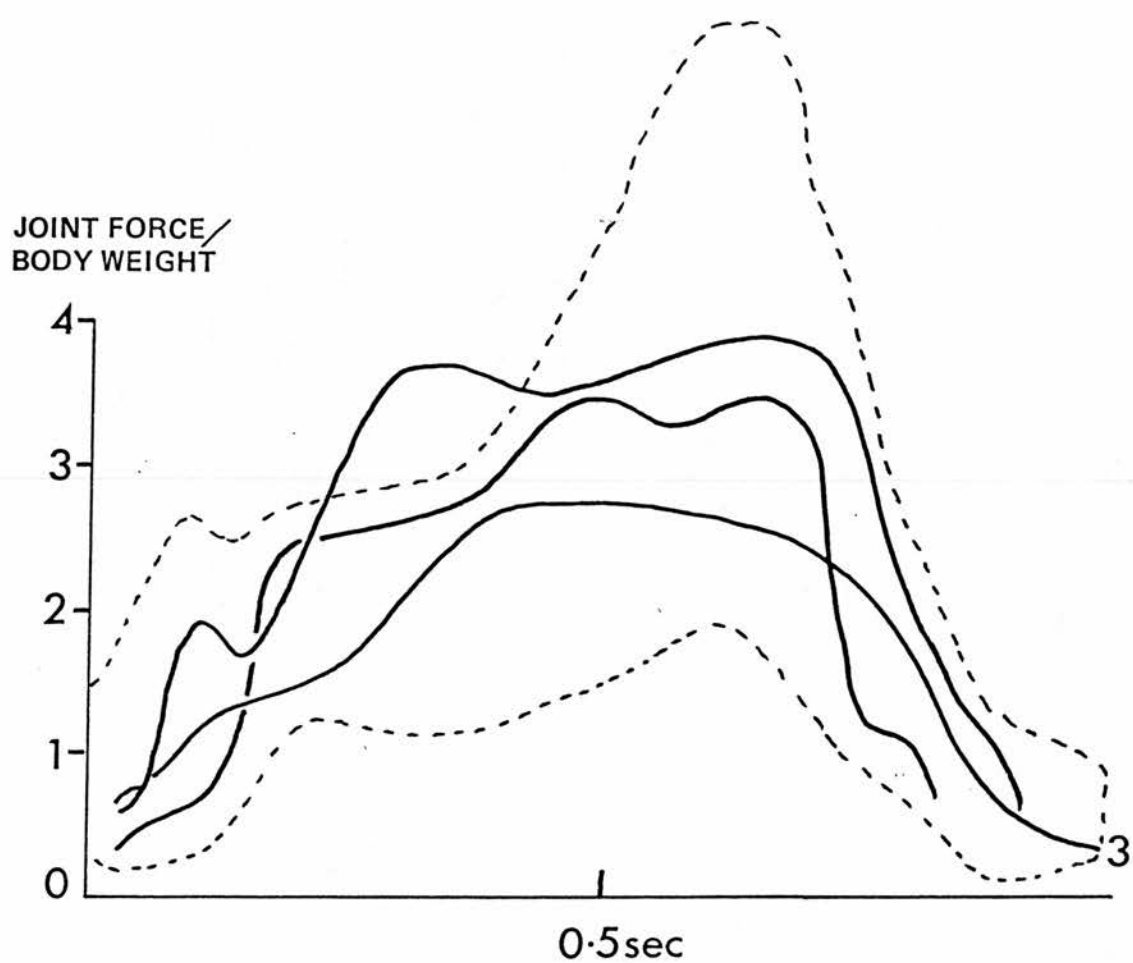


Fig. 1.19 Curves for hip joint loads after total hip replacement lying within the envelope of Paul's values for normal subjects (Paul and McGrouther 1975).

the anatomy was "normal". However, it does indicate that loading across a replaced joint is not likely to be normal and illustrates the fallacy of applying the results of loading values in the normal joint to the endoprosthesis joint, for the purposes of studying prosthetic design.

Tooth (1976) as part of a study investigating the effect of surgical procedures at the joints of the leg on human locomotion studied three female patients two years after Charnley hip arthroplasty for unilateral osteoarthritis of the hip. He tested both lower limbs of the patients during the stance phase of level walking and reported that the load across the prosthetic joint was generally maintained at a lower level than that across the sound joint, and that the pattern of the joint force curve was abnormal for both hips. However, such a comparison was only possible in two of his patients. At the knee joints, in one patient, the load at both knees was similar and in the other there was a marked difference, the load at the knee on the sound side being higher. Conclusions are very difficult to draw from the study of only two patients, but Tooth suggested that a natural extension of his work would be the further investigation of the effects of total hip replacement upon the other joints of both legs. The work to be described in this thesis is the continuation of Tooth's study.

1.VII THE INFLUENCE OF ANATOMICAL FACTORS AT THE HIP
UPON GAIT

Most orthopaedic surgeons are aware of the work of Pauwels in Aachen (1935) which analysed muscle and joint forces at the hip in the static situation of the one-legged stance. Pauwels considered only the frontal plane (syn. coronal plane) and the abductor muscles. He placed the centre of gravity of the body anterior to the second sacral segment in the midline, and estimated that since the ratio of the body weight lever arm to the abductor muscle lever arm was in the ratio of 3:1 the pull of the abductors must be three times the body weight to maintain equilibrium. Denham (1959) published a theoretical paper concerning the clinical significance of alterations in the length of the body weight lever arms and illustrated the way in which the body's centre of gravity changes with posture. McLeish and Charnley (1970) took account of this when they experimentally estimated abductor muscle and hip joint forces by calculating the position of the centre of gravity in various body attitudes of the one-legged stance. Measurements of anatomy were made on radiographs. The lines of actions of the abductor muscles were determined from the dissection of a specimen and the insertion of wires along the apparent centre of action, determined by muscle fibre orientation. The radiographic expression of these lines was then obtained by x-raying the specimen. Hip joint force was estimated at between 1.8 and 2.7 times body weight.

Charnley (1979) later used the results of this experiment to assess the theoretical effect of the anatomical

and surgical variables pertinent to hip joint replacement upon the loading of the joint. He concluded that slight medialisation of the acetabular cup, and lateralisation of the greater trochanter will lower the resultant hip joint force, depending upon the offset of the femoral components used. He makes no mention of the effect of vertical changes in the position of the acetabular cup upon muscle force or joint load, although mention is made of its relationship to limb length discrepancy and subsequent dislocation.

All of the above studies are two-dimensional and ignore the effects of flexor and extensor muscles upon hip joint load. Any figures for hip joint loading will therefore be underestimates. Also, despite the contention of McLeish and Charnley (1970) that the one-legged stance approximates to the stance phase of gait in slow walking, consideration of the work of Paul (1967) and Crowninshield (1978) indicates that this is not the case.

Williams and Svensson (1968) attempted a three-dimensional theoretical analysis of the one-legged stance and obtained values of approximately six times body weight for the hip joint force. This high value and the theory used for its calculation have been severely criticised by Paul (1967) on the grounds that the mechanics were incorrect, the force in the fascia lata (Ft) being unbalanced (Fig. 1.20).

Despite the obvious link between anatomical factors and muscle force, joint load, and therefore hip function,

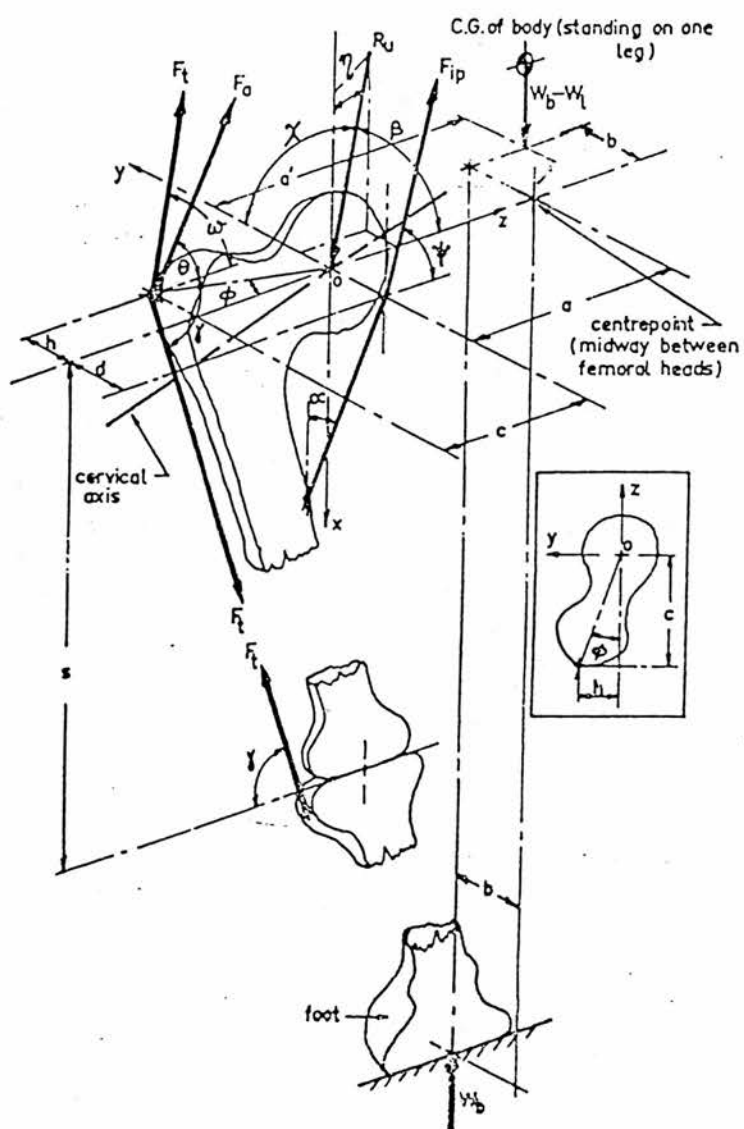


Fig. 1.20 3-Dimensional theoretical analysis of hip joint force during the one-legged stance (Williams and Svensson 1968).

which is apparent from the static studies, surprisingly little attention has been paid to their influence upon the parameters of gait. As already mentioned the Chicago group (Andriacchi and Galante, 1980) had noted the effect of abductor muscle lever arm length on hip moments, but all other work that has been presented has either been kinematic or has investigated loading in a theoretical fashion. All have considered hip joint replacement and no account of anatomical disturbances in degenerative disease of the hip and its relationship to gait parameters has been published.

All of the kinematic data emanates from the Milwaukee group. Gore, Murray, Gardner, and Sepic (1977) related the radiographic positioning of a Muller total hip replacement in 52 patients with no other joint disease to the kinematic data of gait and maximum isometric abductor muscle moment. They showed that, as a group, these patients had the centre of the femoral head 1.2 cm more superior and 0.3 cm medial to that of the contralateral side. The greater trochanter was 0.8 cm further from the femoral head and 1.1 cm more distal. There was some correlation between the distal position of the greater trochanter and greater hip abduction muscle strength but lateral displacement of the greater trochanter and medialisation of the prosthetic head did not correlate with abductor muscle strength. The medial placement of the head, although theoretically advantageous for function (according to Pauwels) was always associated with superior placement of the head and it was felt that the superior

placement correlated significantly with poorer function because of its unfavourable effect upon the length-tension relationship of the abductor muscles. Muscle strength estimations in this study were based upon maximum isometric contractions, a situation rarely encountered in level walking. Therefore, the relevance of correlation of this factor to parameters of gait must be in doubt.

Ranges of motion changes in this study were small and differences between the male and female patients were apparent. Both men and women showed statistically significant correlations of smaller excursions into flexion with a more superior position of the prosthetic head, and shortening of the limb. There were greater excursions into flexion in men with increased neck length, and in women with greater anteversion of the prosthetic femoral component.

This approach was extended to the comparison of different designs of hip arthroplasty, the influence of trochanteric osteotomy and the surgical approach used.

Thirty-five patients with Charnley arthroplasties and 54 patients with Muller arthroplasties, all with trochanteric osteotomy and reattachment were studied pre-operatively and six and twenty-four months post-operatively (Murray et al, 1979). The groups were comparable with regard to gait parameters pre-operatively and two years post-operatively no significant differences were found between them. However, the patients with Charnley arthroplasties showed more improvement in maximal isometric hip abductor muscle moment from before to two years post-operatively, but the difference was statistically

significant only for the women. Mention has already been made of the doubtful significance of this parameter.

A study of 82 patients with Muller arthroplasty, 41 with trochanteric osteotomy and 41 without, was conducted in the same way (Murray et al, 1981b). After taking into account differences in the pre-operative groups it was concluded that there was no long term functional advantage or disadvantage to be derived from satisfactory trochanteric osteotomy. In the 75% of patients with radiographic measurements available there was significantly greater centralisation of the prosthetic femoral head relative to the contralateral femoral head in the group with trochanteric osteotomy. However no account was taken of component positioning differences between the two groups in the analysis of the gait data.

In order to investigate the influence of the surgical approach used upon the positioning of the components of the arthroplasty and the kinematic parameters of gait, Gore and his colleagues (1982) studied 93 patients who had undergone Muller arthroplasty without trochanteric osteotomy. Fifty-two of these patients were operated on through a posterior approach and the remainder through an antero-lateral approach. Patients operated on with the posterior approach had more superior placement of the femoral head but less femoral component anteversion, and longer neck lengths, resulting in more lateral and distal placement of the greater trochanter than those operated upon using the antero-lateral approach. The differences in function that were statistically significant related to component positioning

and were the same as those in the earlier study (Gore et al, 1977) (vide supra).

The parameters of gait which correlated with the surgical approach rather than component position were more internal rotation and more normal abductor muscle strength after a posterior approach was used and more external rotation where an anterolateral approach was used. These features correspond to the anatomical structures divided during each surgical approach.

All of the above studies are valuable contributions to the understanding of the influence of total hip replacement surgery upon gait but without the measurements of kinetic factors it is not possible to determine how the observed changes in the gait result from the observed anatomical changes.

Wiesman and his colleagues in Boston (1978) investigated the effect of trochanteric osteotomy upon hip function by studying twelve patients with bilateral hip replacement, one with, and one without trochanteric osteotomy. The study involved clinical assessment including a Harris score rating, radiographic measurement of "biomechanically relevant parameters", and force plate analysis. The analysis of hip joint force was two-dimensional and based upon the single-legged stance. There were no significant biomechanical differences between the two groups but the changes in length of abductor muscles measured from the radiographs from pre- to post-operatively were significantly different, restoration of the length-tension relationship of the abductors, by either surgical

approach, correlating highly with the patients' preferred hip. They concluded that pre-operative changes in abductor length secondary to joint destruction and intra-operative changes related to the size and positions of the prosthetic components would be a more appropriate way of deciding the position of reattachment of the greater trochanter than the simple and frequently erroneous assumption that more distal reattachment will increase the abductor lever arm. This work is subject to the same criticisms as the work of Pauwels being based upon Pauwelian theory of the one-legged stance and only considering the hip joint in two dimensions. Additionally only ground reaction forces were studied and correlations were based on subjective data i.e. the patient's preferred hip.

Sophisticated theoretical analyses of the influence of anatomical factors upon joint loads after hip joint reconstruction have been performed in Iowa. Bartel and Johnston (1969) studied the cup arthroplasty using a computer programme which indicated the optimal position, based upon the criteria that the pressure across the hip joint should be as small as possible and the force required in the abductor muscle to maintain pelvic stability should be as low as possible. The most important criterion was shown to be the minimisation of abductor force and to achieve this the acetabulum needed to be moved medially and inferiorly and the tip of the greater trochanter required to be moved laterally. This was a static analysis in two dimensions and as such ignored the effects of the flexor

and extensor muscles and it was not related to the changes that occur in muscle activity during gait.

Applying a similar approach to total hip replacement but using a three-dimensional analysis, Johnston, Brand and Crowninshield (1979) studied the effect of location of hip joint centre, trochanteric transfer, valgus placement of the femoral component and the neck stem angle of the prosthesis. They used kinematic and kinetic data obtained from gait analysis using visible LED's for a 27-year-old normal subject in conjunction with a complex muscle model (Crowninshield, 1978) to determine the forces about the hip joint, during level walking, stair climbing and rising from a chair. Anatomical changes were made to the muscle model and ground reaction forces and limb movements were assumed not to change. The conclusions were that (1) in order to reduce the loads at the hip joint, the joint centre should be moved as far medially, inferiorly and anteriorly as is anatomically possible. (2) Moving the greater trochanter laterally is also mechanically desirable but less significant and (3) a short prosthetic neck length and femoral shaft-prosthetic neck angle of 130 degrees appear to be optimum, the joint contact force being increased with higher angles, although not being affected by neck length. The assumptions made in this study, particularly that normal limb motion is seen after arthroplasty, limit its validity but it clearly indicates the importance of component positioning upon hip function. The approach avoids the need for considering multiple interactive variables extant in any given patient with a joint replacement, but it does ignore the wider effects of the positioning of arthroplasty on the other lower limb joints.

1.VIII STUDIES OF GAIT AFTER RESURFACING ARTHROPLASTY OF THE HIP

The author knows of only one relevant publication in this area. Rosenberg et al (1982) in Chicago have studied the gait of ten patients with successful resurfacing arthroplasty of the hip joint and compared the results with those from ten patients with successful total hip replacement. Patterns of motion at the hip, knee and ankle were normal in both groups with total hip replacement patients having a range nearly that of a group of age matched normal subjects, whilst those with resurfacing used a very much smaller range of motion during gait. This group also showed an increased cadence and decreased stride length when compared with normal subjects. Hip moment patterns and peak external forces for the total hip replacement group were within normal limits but the resurfacing patients showed higher adductor and lower abductor moments than either normals or total hip arthroplasty patients. The moments in the transverse and saggital plane were not significantly different from normal.

These findings were felt to indicate dynamic realignment of the trunk over the supporting limb as in the antalgic gait, but none of these patients limped. The possibility of this moment change and the movement pattern being related to a desire to minimise local stresses within the joint or at the proshtesis bone interface was suggested. The methods used for this study have been described in relation to total hip replacement studies and are sound. It would be of interest to know of the joint loads across this type of replacement and

the direction of the resultant force in relation to the fixation of the femoral cup particularly since the pattern of movement and moments reported suggests the possibility of instability at the reconstructed hip joint.

1. IX STUDIES OF GAIT AFTER EXCISION ARTHROPLASTY
 OF THE HIP JOINT

In this procedure the head and neck of the hip joint is resected (Fig. 1.21). It was initially used in the management of tuberculous arthritis (Girdlestone, 1928) but is now most commonly seen as the result of the removal of the components of a failed total hip replacement.

The author knows of no communications concerning the biomechanical analysis of gait in this group of patients. Work assessing the clinical function of patients after this procedure (Haw and Gray, 1976; Murray, Lucas and Inman, 1964) describes a severe short leg limp with a high proportion of patients requiring some form of walking aid.

GIRDLESTONE ARTHROPLASTY

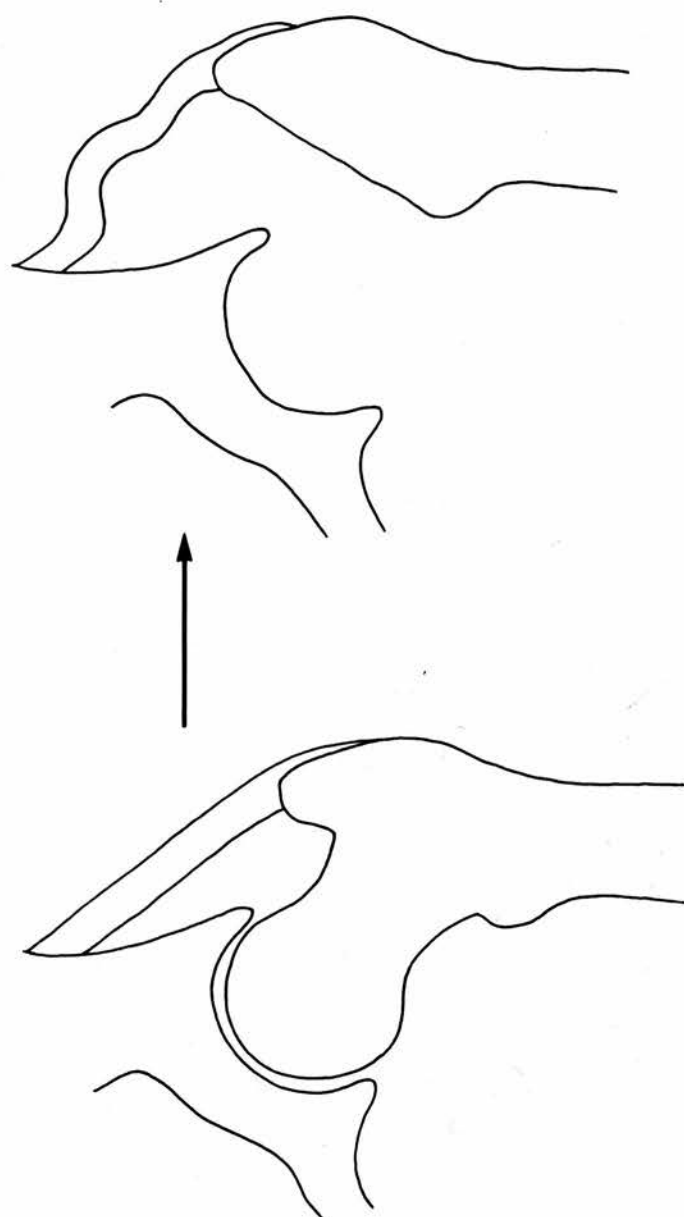


Fig. 1.21 Diagrammatic representation of excision arthroplasty of the hip joint.

CHAPTER 2

MATERIALS AND METHODS

I PATIENTS

1. Total Hip Replacement
2. Excision Arthroplasty

II EQUIPMENT

III CALIBRATION OF EQUIPMENT

IV EXPERIMENTAL PROCEDURE

1. Preparations
2. Marker Placement
3. Conduct of a Test

V DATA HANDLING

VI DATA INTERPRETATION

2.I PATIENTS

Two groups of patients were studied.

1. Total Hip Arthroplasty

Sixteen patients were selected from those admitted to the Professorial Unit of Gartnavel General Hospital, Glasgow, for total hip replacement surgery. The criteria for selection were :-

- i. degenerative disease of one hip joint and no clinical evidence of disease in any other joint of the other lower limb.
- ii. the ability to walk 100 metres unaided, or with the use of one stick, it being found early in the course of the MRC funded study of which the work for this thesis formed a part, that patients who could not meet this criterion were difficult to test and were rapidly fatigued.

There were six men and ten women (Table 2.1). The mean age of the group was 58 years (28 to 74 years). Fourteen patients were suffering from osteoarthritis and two were diagnosed as having a polyarthropathy. These last patients had other joints involved in the upper limbs only.

All patients underwent a full clinical examination and details relevant to hip function were recorded on a Harris hip proforma which is the routine procedure for all hip replacement patients in the unit concerned. A clinical assessment of gait formed part of this examination and was used to aid the selection of patients for study.

An attempt was made to obtain standard radiographs of the pelvis pre-operatively, the radiographers being

PATIENT	AGE	SEX	WEIGHT kg.	HEIGHT m.	DIAGNOSIS	AFFECTED SIDE	OPERATIVE PROCEDURE
1	64	M	85.7	1.75	O.A.	Right	C.A.D.Muller
2	63	F	62.1	1.59	O.A. Cong. coxa vara	Left	Charnley
3	33	F	49.5	1.64	Mixed connective tissue syndrome	Left	C.A.D.Muller
4	35	M	110.5	1.88	Avascular necrosis Thompson Arthroplasty	Right	Charnley
5	51	F	45.4	1.65	O.A.	Right	C.A.D.Muller
6	35	M	100.0	1.79	Avascular necrosis Post. Dislocation	Right	Charnley
7	56	M	77.6	1.72	Caisson's Disease C.A.D.Endoprosthesis	Right	Charnley
8	47	F	79.0	1.59	O.A.	Right	C.A.D.Muller
9	60	F	56.3	1.53	O.A.Intertroch- anteric osteotomy	Left	Charnley
10	73	F	56.5	1.59	O.A.	Right	Charnley
11	65	M	70.7	1.69	O.A.	Left	C.A.D.Muller
12	70	F	71.2	1.58	O.A.	Left	C.A.D.Muller
13	58	F	69.5	1.66	O.A.	Right	C.A.D.Muller
14	56	M	97.8	1.82	O.A.	Left	Charnley
15	59	F	66.9	1.63	O.A.	Right	C.A.D.Muller
16	33	F	62.7	1.65	Intertroch.osteo- tomy. O.A. Old C.D.H.	Left	Charnley

CHAPTER 2 TABLE 1. TOTAL HIP REPLACEMENT PATIENTS

Notes: 1. Patient 11 was not studied post-operatively.

2. Patient 25 from the Girdlestone Group will be considered with the
H.J.R. post-operative group

instructed to centre the X-ray beam over the symphysis pubis, use a gun to film distance of 100 cm, and ensure that the patients' lower limbs were in line with the trunk with the patellae pointing anteriorly. Despite these instructions, in common with other investigators, (Gore et al, 1977, Weisman et al, 1977) a great variation in the standard of these films was noted. Lateral views of the affected hip were performed using the Manfredi shoot-through technique. In addition to these "standard" films, full length standing radiographs of both lower limbs were performed (see Fig. 2.1) to enable the anatomical relationships of the femoral axis to be determined radiographically (Chapter 3).

Eight patients were treated with a Charnley arthroplasty (Thackray) using the technique described by Charnley (1979) which involves osteotomy of the greater trochanter and its subsequent reattachment. The 3-wire technique for trochanteric reattachment was used in every case (Boardman et al, 1978). The remaining nine patients were treated using a CAD (Computer Assisted Design) Müller arthroplasty (Howmedica) which was inserted through an anterolateral approach as described by Müller (1978). The other major difference in technique was that the acetabulum was prepared with gouges alone in all but one of the patients with a CAD Müller arthroplasty whereas all patients treated with a Charnley arthroplasty had their acetabulum prepared using Charnley reamers (Charnley, 1979). The difference in design of the two femoral components is clear from Fig. 2.2 and will be discussed further in Chapter 4.



Fig. 2.1 Full length standing radiograph
(ankles omitted from photograph)



Fig. 2.2 Charnley,,(Left) and
C.A.D. Muller (Right)
femoral prostheses

The post-operative regime was the same for all patients with the exception of active abduction which was avoided in patients with Charnley arthroplasties until the six-week check film had been inspected for signs of union or separation of the greater trochanter. No protection of weight-bearing was used but most patients required to use a Zimmer frame at the beginning of mobilisation on the fourth post-operative day. No specific muscle strengthening exercises were used in any patient. Many left hospital two weeks post-operatively using one stick and all but one of the patients studied had discarded this by one year post-operatively.

Prior to discharge from hospital all patients had "standard" radiographs of the pelvis and a lateral view of the replaced hip performed. These acted as "reference films" for later comparisons. One patient (patient 11), who lived a considerable distance from Glasgow, was not able to attend for biomechanical testing (vide infra) at one year and will therefore only be considered with regard to his pre-operative state.

2. Excision Arthroplasty

Nine patients who had their total hip replacement arthroplasties removed because of infection between three and eight years prior to study were selected on the basis of their ability to walk unaided or with one stick over a distance of 100 metres. This group included patients with other joint problems (see Table 2.2) and they were tested biomechanically on only one occasion.

CHAPTER 2 TABLE 2

GIRDLESTONE EXCISION ARTHROPLASTY PATIENTS

Patient	Age Yrs.	Sex	Wt. Kg.	Ht. m.	Affected side	Other Involved Lower Limb Joints	Time Since Excision (Years)
24	62	M	64.3	1.56	Right	Nil	4
25	55	F	66.0	1.65	Left	Painful Charnley T.H.R. Right Hip	5
26	68	M	67.1	1.66	Right	Nil	8
27	77	F	60.7	1.56	Left	Howse T.H.R. Right Hip	4
28	58	F			Left	Howse T.H.R. Right Hip	4
29	70	M	79.0	1.74	Right	Nil	3
30	71	F	76.1	1.62	Left	O.A. Right Hip	5
31	48	F	51.1	1.52	Left	Nil	3
32	64	F	67.0	1.73	Right	Nil	4

They were assessed clinically at the time of their test and a Harris hip proforma was completed. Since all were still under review in the clinics of the University Department of Orthopaedic Surgery, Glasgow, radiographs were available but no full length standing films were obtained. One of these patients (25) later underwent conversion to a further total hip replacement of Charnley type and is grouped with the total hip arthroplasty group for consideration of her post-operative state.

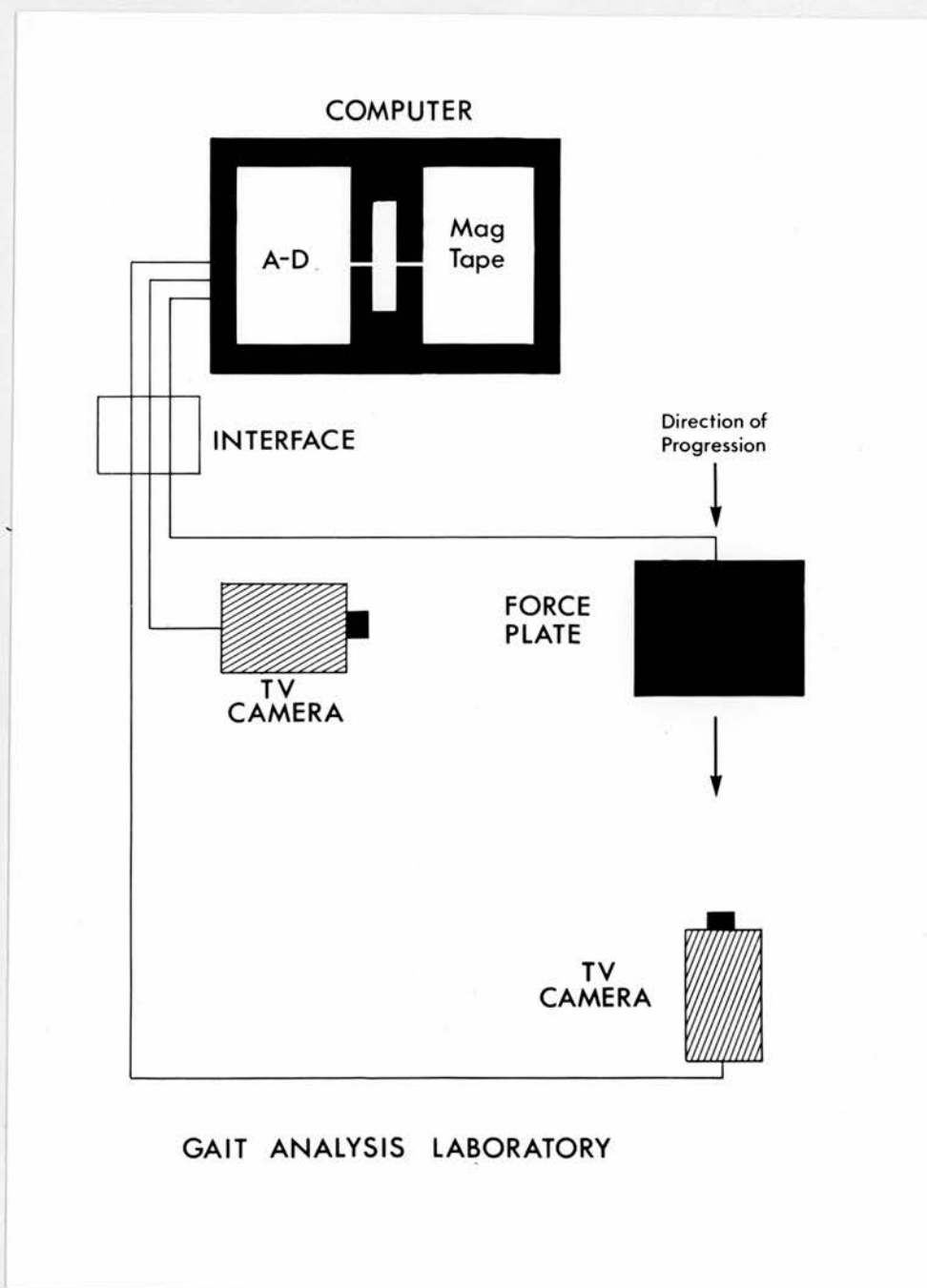


Fig. 2.3 Plan of Gait Analysis Laboratory

2.II EQUIPMENT

The gait laboratory of the Bioengineering Unit (Fig. 2.3) is in the basement of the Wolfson Centre. It is a large area, free from encumbrances, and with a high ceiling. Lighting is a mixture of natural light and mercury vapour lamps. The floor is covered with a non-slip P.V.C. and the 50 feet walkway forms an integral part of the floor.

Patients' movements during gait were recorded from two of three video cameras, (Vidicon Tubes), one placed to view along the axis of progression and two others perpendicular to this angle of view to record data from the right or left lower limbs. A halogen lamp was placed to shine along the optical axis of each camera in order to enhance reflections from the markers worn by the patient (vide infra).

Floor reaction forces were measured using one of two piezo-electric force plates (Kistler Instruments AG, Switzerland) which were set into the floor of the laboratory and covered with the same P.V.C. This force plate, the design of which was described in Chapter 1.III 1, measured the force applied to it as three orthogonal components together with the three vector components of moment relative to the axes of the platform. Data collected from the force plate at 50 Hz was stored in a Digital PDP - 12 computer. The method by which this was done can be found in Tooth (1976). Synchronous recordings of motion from the television cameras (at 50 Hz) were also passed to the computer using an interface described by Jarrett et al (1976).

2.III CALIBRATION OF EQUIPMENT

T.V. System

A calibration board with retroreflective markers in a diamond pattern was used (Fig. 2.4). The distance between each marker was precisely known. By placing this board over the force plate and exposing it to the front and side cameras, the displacement measured as units in the computer store were converted to metres and the centre of the reference axis system at the force plate was defined.

Force Plate

The accuracy of the force plate was checked monthly by the use of a pointed rod which was loaded with a known weight. This load could then be applied at given points on the force plate and the output for load and centre of pressure checked.

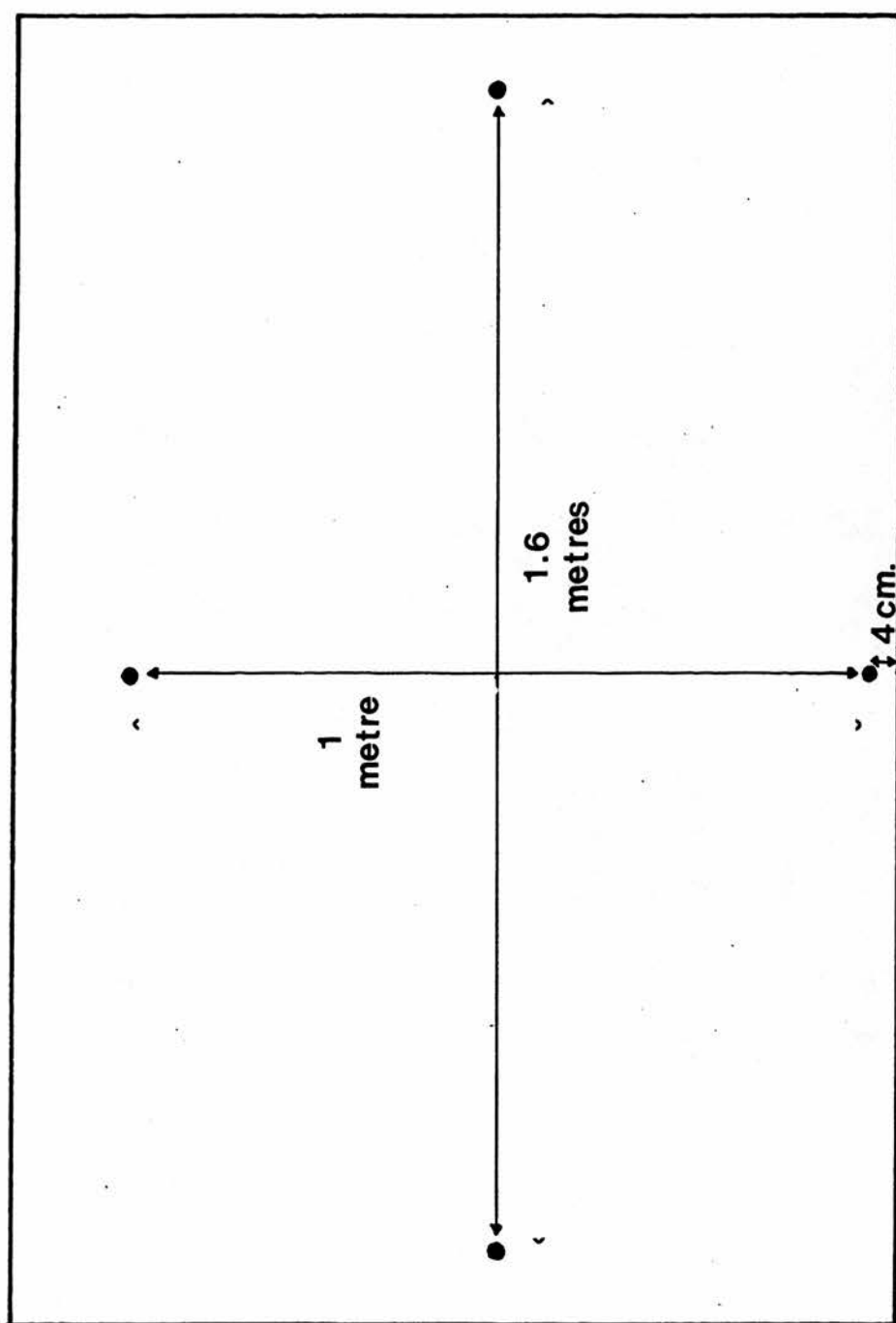


Fig. 2.4 Diagram of calibration board
for T.V. System

2.IV EXPERIMENTAL PROCEDURE

1. Preparations

The cameras were switched on at least one hour before the test was due to begin to allow the Vidicon tubes to warm up. Immediately prior to the beginning of the test the cameras were calibrated, as described above, and then apertures were adjusted to allow only reflections from the markers to be detected.

Patients changed into black shorts "tailored" to allow placement of markers on the pelvis (see Fig. 2.5). Because of the camera system used, white clothing produced reflections with consequent noise in the television signal. If patient was wearing a white shirt or blouse this was removed or replaced by a dark-coloured T-shirt. Watches were removed for the same reason. All tests were performed in outdoor shoes and the patient was weighed with and without shoes prior to testing. A note of the height was also taken.

2. Marker Placement

The markers consisted of three 1 cm diameter spheres mounted on short balsa wood stalks, a similar marker with a 10 cm stalk, two 1 cm diameter hemispheres, and six 1 cm diameter bevelled, circular markers. All were covered with retroreflective tape (3M) and were secured to the patient with double-sided adhesive tape.

The markers were placed at sites where movement of the skin relative to underlying bone was likely to be small. The pelvis was defined by one hemispherical marker over



Fig. 2.5 Mode of Dress used During Tests.

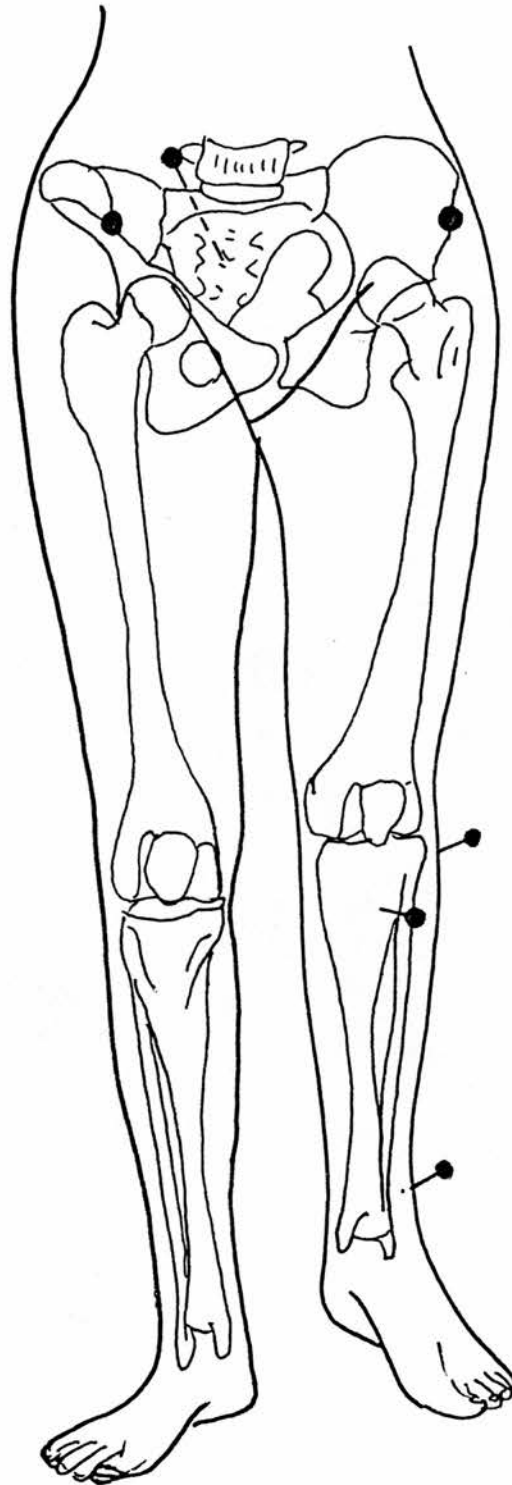


Fig. 2.6 Placement of retroreflective markers.

each anterior superior iliac spine and the tail marker, on its long stalk, over the midline of the sacrum at the level of the lumbosacral junction. The lower limbs were studied individually and markers on short stalks were placed over the fibular head, the lateral malleolus and the anterior subcutaneous border of the tibia (see Fig. 2.6). At the conclusion of the tests on each limb, additional bevelled circular markers were placed over what was estimated to be the hip, knee and ankle joint centres in the anteroposterior and medio-lateral planed. This information was fed into the computer with the patient standing in front of the cameras and over the centre of the reference axes. In this way the amount of information which requires to be processed was limited, simplifying later data analysis.

3. Conduct of a Test

With the exception of some pre-operative patients and some patients in the excision arthroplasty group, all tests were carried out without the use of a walking aid. The patient was asked to walk towards the front camera and was not told of the existence of the force plate, although its site in the floor was obvious. Several conditioning walks were performed to enable a starting position to be selected from which the patient would achieve a single strike with one foot entirely on the force plate. This proved to be extremely difficult in the pre-operative group of patients because of the irregularity of successive step lengths. Further difficulty was caused by the presence of a short step length or a narrow based gait and this had to be overcome by making changes in the line of walking.

Once a suitable starting point seemed to have been found motion and displacement data were recorded on to magnetic tape in the PDP 12 computer.

In the pre-operative patients in particular it was not uncommon for the arm on the side of the leg under study to obscure the pelvic marker on that side during the test. The computer programmes used (vide infra) enabled up to 10 missing points in the movement data to be predicted and inserted but if more points than this were missing it was necessary to ask the patient to walk with the offending arm flexed across his or her chest.

An attempt was made to obtain data for at least two successful test walks for each lower limb but this was not always possible pre-operatively.

A major advantage of the television-computer system over previous cine-film studies is that the movement and ground reaction force data can be replayed within two minutes of the test allowing deficiencies to be recognised and only acceptable tests to be recorded on to the magnetic tape. This is an important factor when studying pathological conditions. The mean number of test walks to obtain two satisfactory sets of records for each limb was 20.

Pre-operative tests occupied between 45 and 75 minutes but post-operative tests were shorter (30 to 45 minutes).

At the completion of the tests on each limb anthropomorphic measurements of the pelvis and the segments of the limb (see Figure 2.7) were recorded on a standard record sheet.

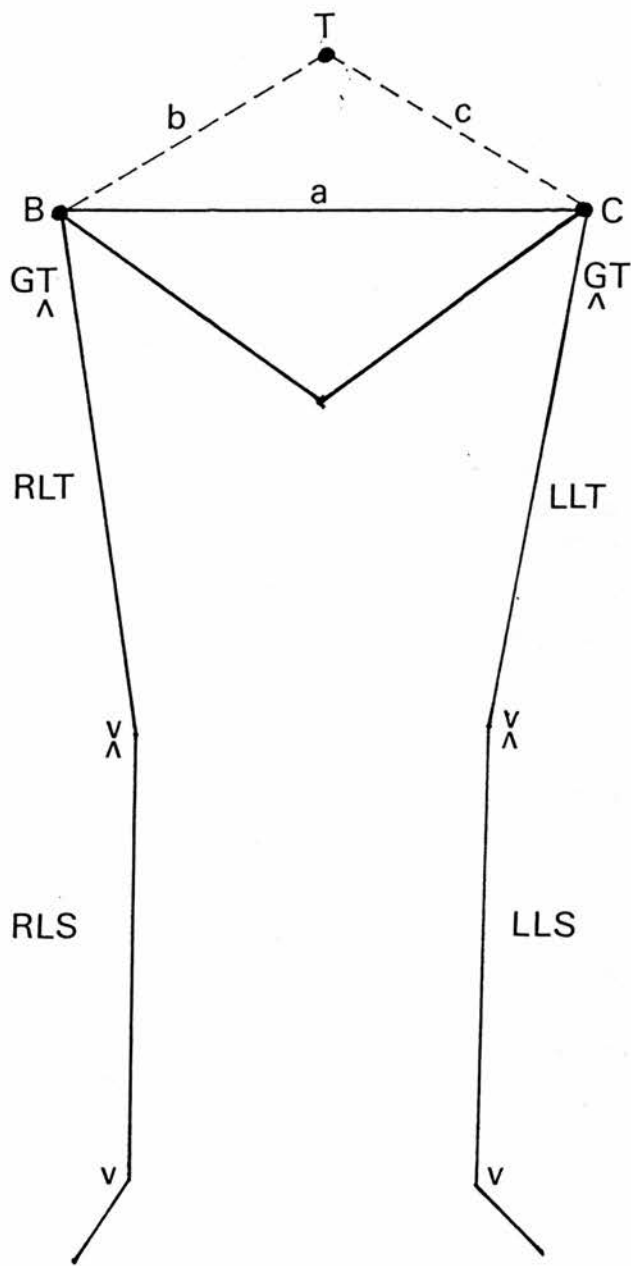


Fig. 2.7 Diagram of Anthropomorphic measurements.

2.V DATA HANDLING

The theoretical basis for data reduction and analysis will be discussed in Chapter 3. Ground reaction force and movement data recorded on to magnetic tape was replayed and the duration of stance phase was determined from the force plate data which was then recorded on to punched tape. Movement data for the same period was inspected, resulting in an analysis of stance phase alone. Using programme EDITOR (written by M. Jordan, 1976) missing motion data points were inserted and the coordinates of each marker at each point in stance phase were noted and transferred to punched tape in a set order. The programme allowed this calculation to be performed automatically once the marker had been identified to it. Thus motion and force plate data were synchronised.

A number of basic parameters (e.g. body mass and segment lengths) were spliced on to the punched tape before input into the ICL 1904S computer which carried out the theoretical procedures to be described in Chapter 3.

The method of data handling represents a significant advance over that of Paul (1967), McGrouther (1974) and Tooth (1967) whose time for data processing was in excess of six hours. Using the described approach the time to test patients and prepare data was between 2½ and 4½ hours.

2.VI DATA INTERPRETATION

Output from the ICL 1904S was in the form of raw numerical data and of the various parameters (e.g. movements, moments and forces), plotted against stance phase duration. The duration of stance phase was normalised to allow comparison of data from different patients and for the same reason forces were expressed in multiples of body weight rather than in absolute units.

The maximal values quoted in Chapter 4 and the phasing of these maxima were measured from the curves.

Multiple linear regression analyses and other statistical tests were performed using the Statistical Package for the Social Sciences (S.P.S.S.).

CHAPTER 3

THEORETICAL ANALYSIS

- I INTRODUCTION
- II AXIS SYSTEMS
- III MEASUREMENTS OF MOVEMENTS AND CORRECTION
OF PARALLAX ERRORS
- IV CALCULATION OF THE UNKNOWN COORDINATE
- V LOCATION OF JOINT CENTRES
 - 1. Hip Joint
 - 2. Knee Joint
- VI RELATIONSHIP BETWEEN AXIS SYSTEMS
- VII RESULTANT FORCES AND MOMENTS
- VIII MUSCLE FORCES
 - 1. Muscle Model
 - a. Muscle grouping
 - b. Muscle attachments and
lines of action
 - 2. Calculation of muscle forces
 - 3. Knee Musculature
 - 4. Hip Musculature
- IX JOINT FORCES
 - 1. Knee Joint Force
 - a) Cruciate ligaments
 - b) Collateral ligaments
 - 2. Hip Joint Force

3.I INTRODUCTION

The fundamental principle of the method of gait assessment used in this study is the measurement of the ground-foot reaction forces using a force plate, and the relation of these forces to the movements of the limb to allow the calculation of intersegmental forces and moments. This means that only the stance phase of gait will be studied (it is during this phase that joint loading is maximal). The gravitational and inertial effects will not be taken into account in this study. As mentioned (1.III 1.) Bresler and Frankel (1950) felt that the contribution of these effects to the resultant joint moments was small, not exceeding 10% of the load actions transmitted between the peak loading times in the gait cycle (12% and 50%. Paul, 1967). A more recent study by Mikosz and his colleagues in Chicago (1982) has calculated resultant forces and moments at the hip and knee with and without segmental inertia. They concluded that the quasi-static forces (calculated from the position data and floor reaction forces) constituted the dominant portion of the stance phase loading at the hip and knee joints, and that the inertial contributions of the limb segments to joint loads during stance phase were secondary and sometimes negligible.

Once the resultant intersegmental forces and moments have been determined, they are balanced by forces distributed between muscles, ligaments and articular surfaces according to a muscle model (1.IV.3.). This process requires

information about the positions of origins and insertions of muscles which is usually obtained from cadaveric dissections.

The basic principles of mechanics necessary for these calculations will be outlined in the remaining sections of this chapter. The principles are not complex but they are repetitive and the analyses have to cope with a large amount of raw data. The use of a computer is therefore essential if more than one or two individuals are to be investigated. The computer programmes used in this study were a modification of Tooth's programmes (1976) made by Dr. T.R.M. Brown in the Department of Bioengineering at the University of Strathclyde. They were not the work of this author and are therefore not included with this thesis.

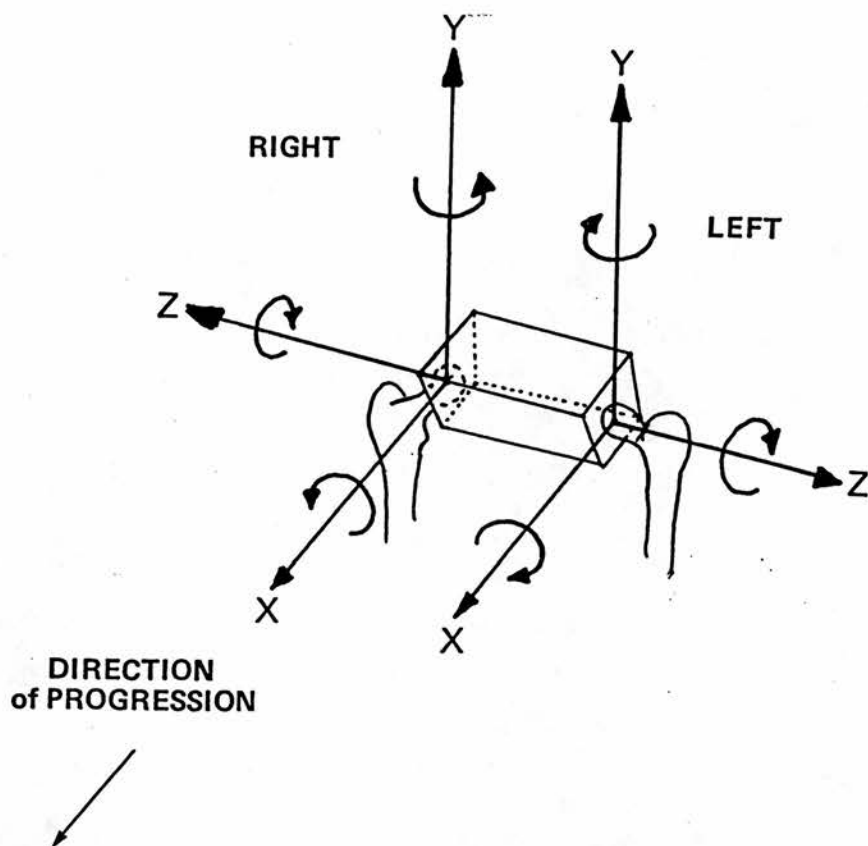


Fig. 3.1 Co-ordinate System at hip joints
and sign convention for moments.

3.II AXIS SYSTEMS

The coordinate systems used for this analysis (Fig. 3.1) are orthogonal and with the subject standing in the normal erect posture have the X-axis positive in the line of progression, and parallel to the floor; the Y-axis vertical and positive upwards; and the Z-axis parallel to the floor and positive when moving away from each limb in a lateral direction. This unorthodox designation of the Z-axis enables traces of forces and moments from the joints of each limb to be presented with the same sign convention. The convention used for the sign of the direction of the moments is also shown in Figure 3.1.

3.III MEASUREMENT OF MOVEMENTS AND CORRECTION OF PARALLAX ERRORS

The movements quoted in this thesis represent angles measured from the television data, ignoring the effect of segmental rotation, after the correction of parallax errors. Because of the axis system used one set of equations will correct for parallax for either side.

The axes, X,Y and Z (Fig. 3.2) are centred at the origin of the laboratory reference axis system. A point P whose true coordinates are x, y and z will have the apparent coordinates x', y' and z' measured from the T.V. data. From Figure 3.2

$$y = y' - (y' - h) z'/R_z \quad \text{3(1)}$$

$$x = x' - x' z'/R_z \quad \text{3(2)}$$

$$z = z' - z' x'/R_x \quad \text{3(3)}$$

From 3(2) and 3(3)

$$x = x' (1 - z'/R_z) / (1 - x' z'/R_x R_z) \quad \text{3(4)}$$

$$z = z' (1 - x'/R_x) / (1 - x' z'/R_x R_z) \quad \text{3(5)}$$

Where the y coordinate is measured from the front camera

$$y = y' - (y' - h) x'/R_x \quad \text{3(1a)}$$

Using these expressions (3(4), 3(5) and 3(6)) all coordinates are corrected for parallax. The distances Rx and Rz are known from the laboratory layout, and the apparent coordinates are in the planes of the calibration boards.

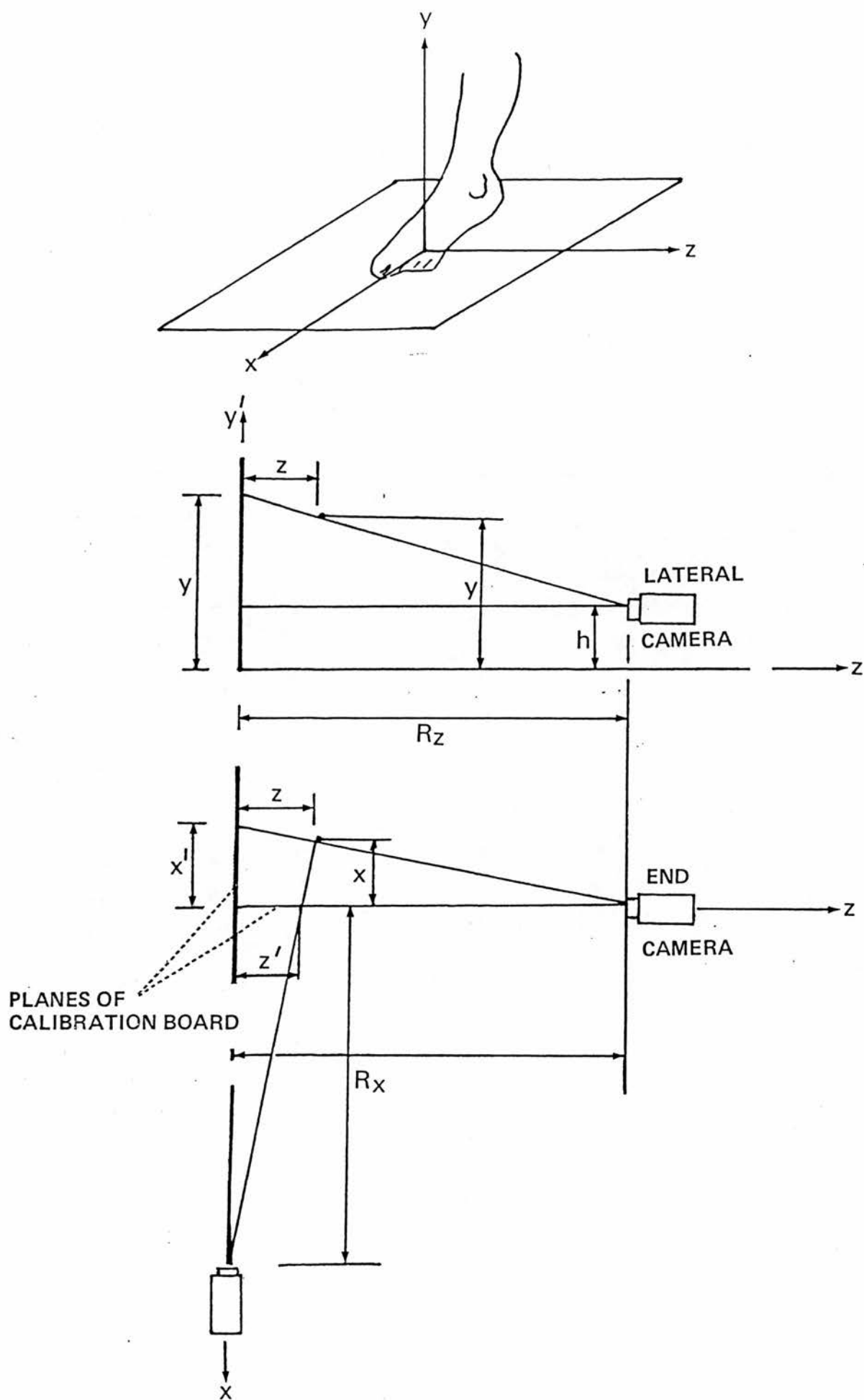


Fig. 3.2 Parallax Correction.

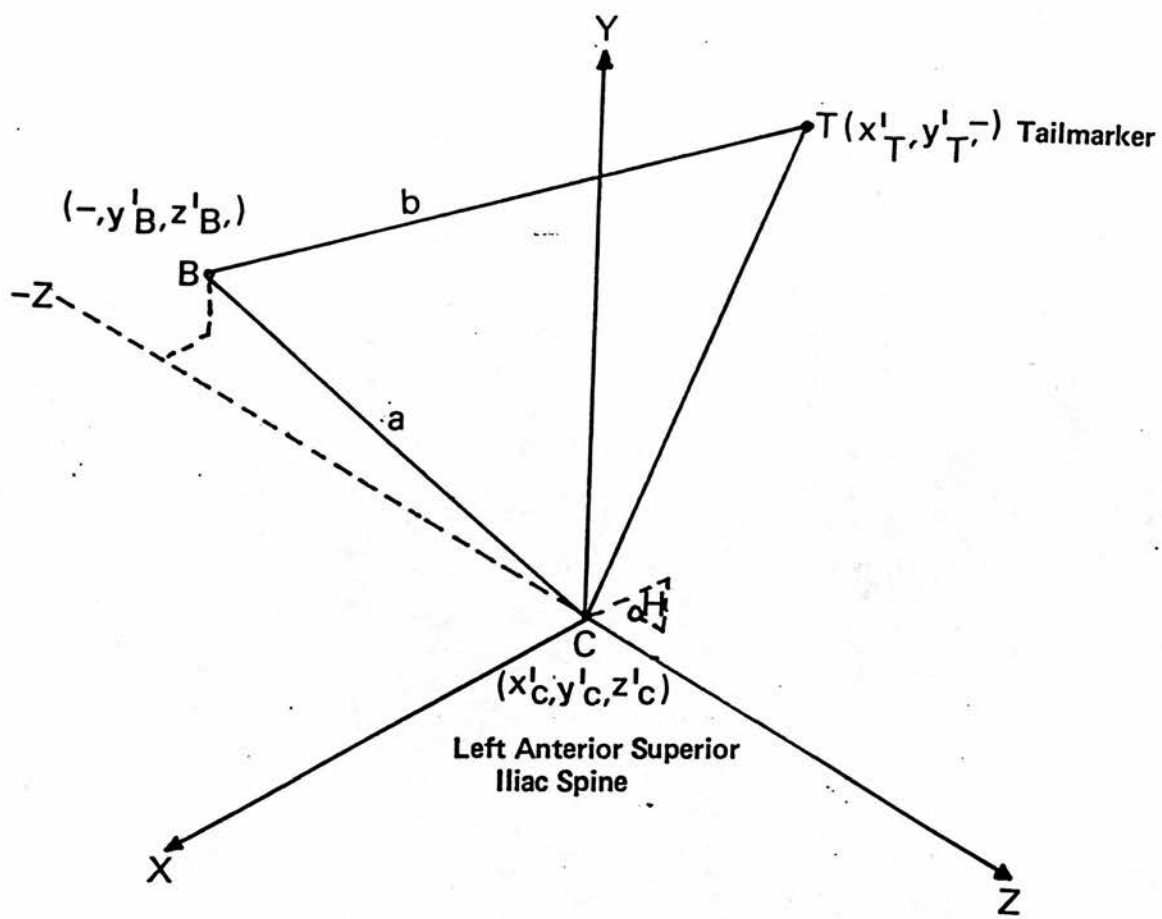


Fig. 3.3 Calculation of the Unknown Co-ordinate.

3.IV CALCULATION OF THE UNKNOWN COORDINATE

All of the markers used can be visualised by both cameras except the tail marker (T) (Fig. 3.3), and the anterior superior iliac spine marker on the side distant from the side camera (B). Therefore, the coordinates of Z_T and X_B are unknown and since X_B is unknown it is not possible to obtain Z_B or y_B , only Z'_B and y'_B .

If it is assumed that the tail marker (T) lies midway between the two anterior superior spine markers (C and B) then :

$$Z_T = \text{approx. } Z'_C - \frac{CB}{2} \quad \text{3(6)}$$

Z'_C is known from the front camera and CB is measured at the time of the test.

Substituting 3(6) into equation 3(4)

$$X_T = X'_T - X'_T \left(Z'_C - \frac{CB}{2} \right) / R_z \quad \text{3(7)}$$

Having determined the true X coordinate for point T the true Y coordinate can be obtained from equation 3(1a) :

$$Y_T = Y'_T - Y'_T - h) X_T / R_z \quad \text{3(8)}$$

Knowing X_T and Y_T , true Z_T can be calculated from :

$$(X_C - X_T)^2 + (Y_C - Y_T)^2 + (Z_C - Z_T)^2 = (C.T)^2 \quad \text{3(9)}$$

For point B, Z'_B and y'_B are known from the front camera but X'_B is unknown.

If it is assumed that:

$$X_B = \text{approx. } X_C \quad \text{_____} \quad 3(10)$$

then by substituting in 3(1a)

$$y_B = y'_B - (y'_B - h) X_C / R_z \quad \text{_____} \quad 3(11)$$

and by substituting 3(10) into 3(3)

$$z_B = z'_B - z'_B X_C / R_x \quad \text{_____} \quad 3(12)$$

true X_B can be obtained from :

$$(X_B - X_T)^2 + (y_B - y_T)^2 + (z_B - z_T)^2 = b^2 \quad \text{_____} \quad 3(13)$$

where b is measured at the time of testing (Fig. 3.3).

This value of X_B can then be used to adjust y'_B and z'_B if necessary.

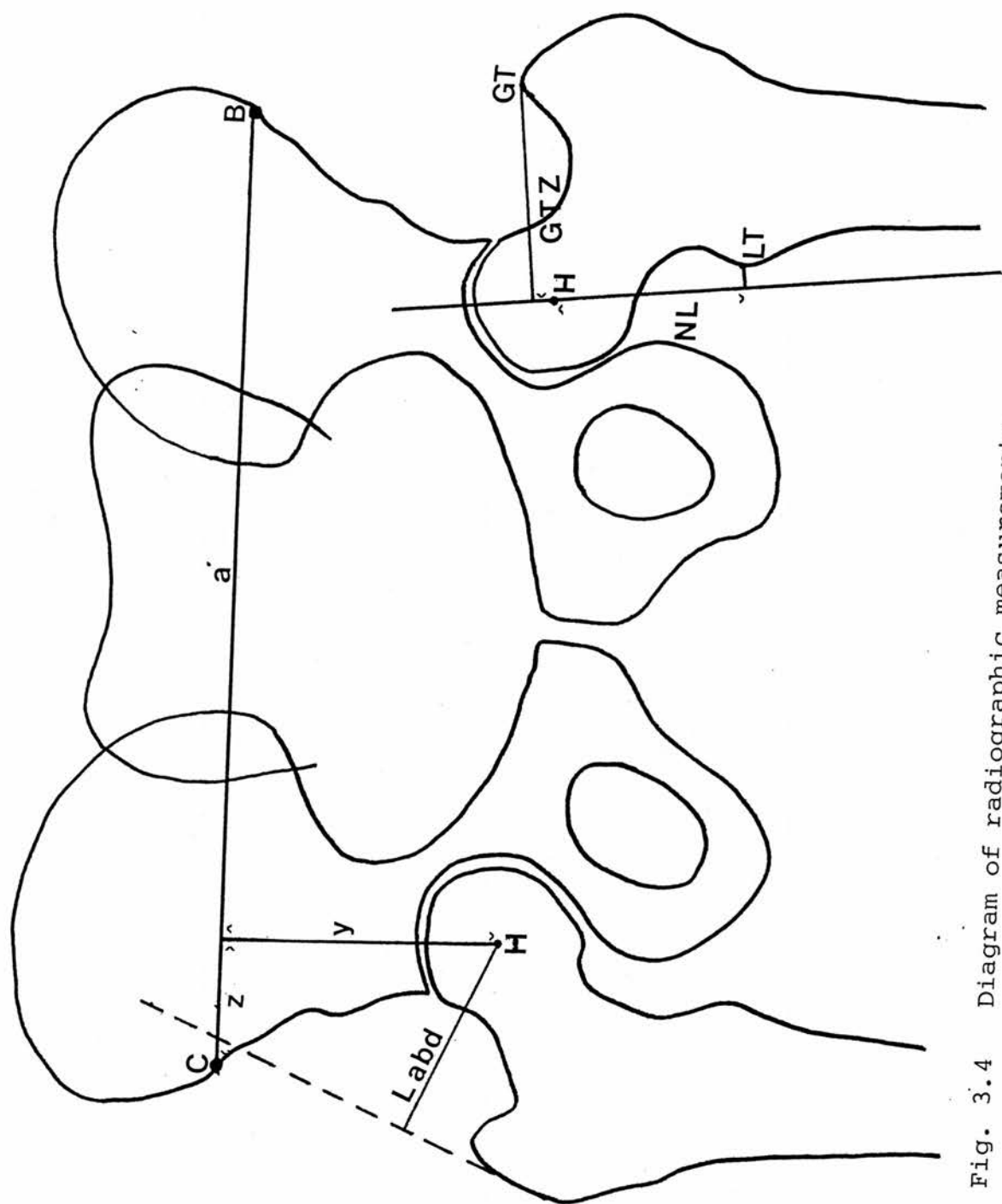


Fig. 3.4 Diagram of radiographic measurements.

3.V LOCATION OF JOINT CENTRES

1. Hip Joint

The Y and Z coordinates of the hip joint centre (H) with respect to the anterior superior iliac spine are measured from standardised antero-posterior radiographs of the pelvis (Chapter 2.I and Figure 3.4). They are corrected for magnification by the factor :

$$\frac{a}{a'} \quad \text{-----} \quad 3(14)$$

where a is the distance CB measured at the time of the test and a' is the distance between the anterior superior iliac spines on the AP radiograph of the pelvis.

The position of the centre of rotation of the hip joint on the AP radiographs was estimated by the use of Moses' rings or, if the femoral head was mis-shapen, by the point of intersection of the perpendiculars to 3 tangents to the femoral head, or the centre of the small triangle formed by these lines. For the excision arthroplasty patients the estimation of the position of the centre of 'rotation' was difficult since the nature of each articulation was not accurately known. However, in order to obtain a point to allow calculations the point midway between the tip of the greater and lesser trochanters was used in each case.

It is recognised that measurements from radiographs are potentially inaccurate because of the inclination of the pelvis and lumbar spine, rotation of the limbs, and the angle of the X-ray beam (Seradge, Nagle and Miller, 1982). Since it was not possible for

this author to supervise every radiograph for this study, not all were of the standard that was desired. This was particularly true for the pre-operative films where abduction, adduction, or rotation contractures at the hip joint not infrequently prevented the radiographers positioning the patient ideally (Chapter 2.I) and where flexion contractures resulted in extreme pelvic tilt in some cases. Post-operatively it was always possible to obtain at least one acceptable radiograph.

These variations were accepted since except in the case of excessive limb rotation, the same factors were relevant for both hips, and the basis for this study is predominantly the comparison of affected and contralateral joints. Where excessive rotatory deformity was present this fact was noted but no attempt was made to correct for it.

Corrections were made for abduction or adduction of the hip when measurements of abductor muscle length or lever arm were being made. The method used for the measurement of the abductor line of action, and lever arm on the radiograph was that described by Charnley (1979) arising from the work of McLeish and Charnley (1970), and the method for measurement of abductor muscle length was that indicated by Weisman et al (1977). The correction was carried out using a goniometer centred on the centre of the femoral head and aligned with the femoral axis and the line joining the joint centre to the tip of the greater trochanter. The goniometer was then rotated until the femoral axis subtended the same angle to the

line joining the anterior superior iliac spines as on the (usually) well aligned contralateral side.

Because of the rarity with which the anterior superior iliac spine was visible on the lateral radiograph of the hip joint the coordinates of the joint centre in the sagittal plane could not be determined from the radiographs. Its antero-posterior position in this plane was therefore calculated using the linear three dimensional proportionality relationships established by Paul (1967) and Poulson (1973).

2. Knee Joint

The centre of rotation of the knee joint was determined by the placing of skin markers during the test, and related to the measurements of the knee taken at the time of testing by the constants of proportionality quoted by Morrison (1967).

3.VI RELATIONSHIP BETWEEN AXIS SYSTEMS

In order to calculate the moments about the hip and knee joints it is necessary to define the relationship between the axis system at the ground (the laboratory reference axis system) and that at each joint. Additionally the relationship between the pelvic and femoral axis systems and the tibial axis system must be known to calculate the muscle and ligament forces at the knee.

The tibial axis system is defined with the foot pointing forwards, that is along the X-axis of progression. The Y and X axes are then mutually perpendicular to this axis. The femoral axis system (Fig. 3.5) has its Y axis defined by the line HA (Harrington, 1974). From the cross product of this axis and the tibial X-axis the femoral Z-axis is defined. This can then be used to determine the true X-axis at the femur.

The pelvic axis system is centred on the anterior superior iliac spine of the side under consideration (C) and is assumed to be parallel to the ground when the subject is in the normal erect posture.

Use of this system permits segmental rotations to be accounted for in the calculation of forces and moments and in the subsequent calculation of the joint force.

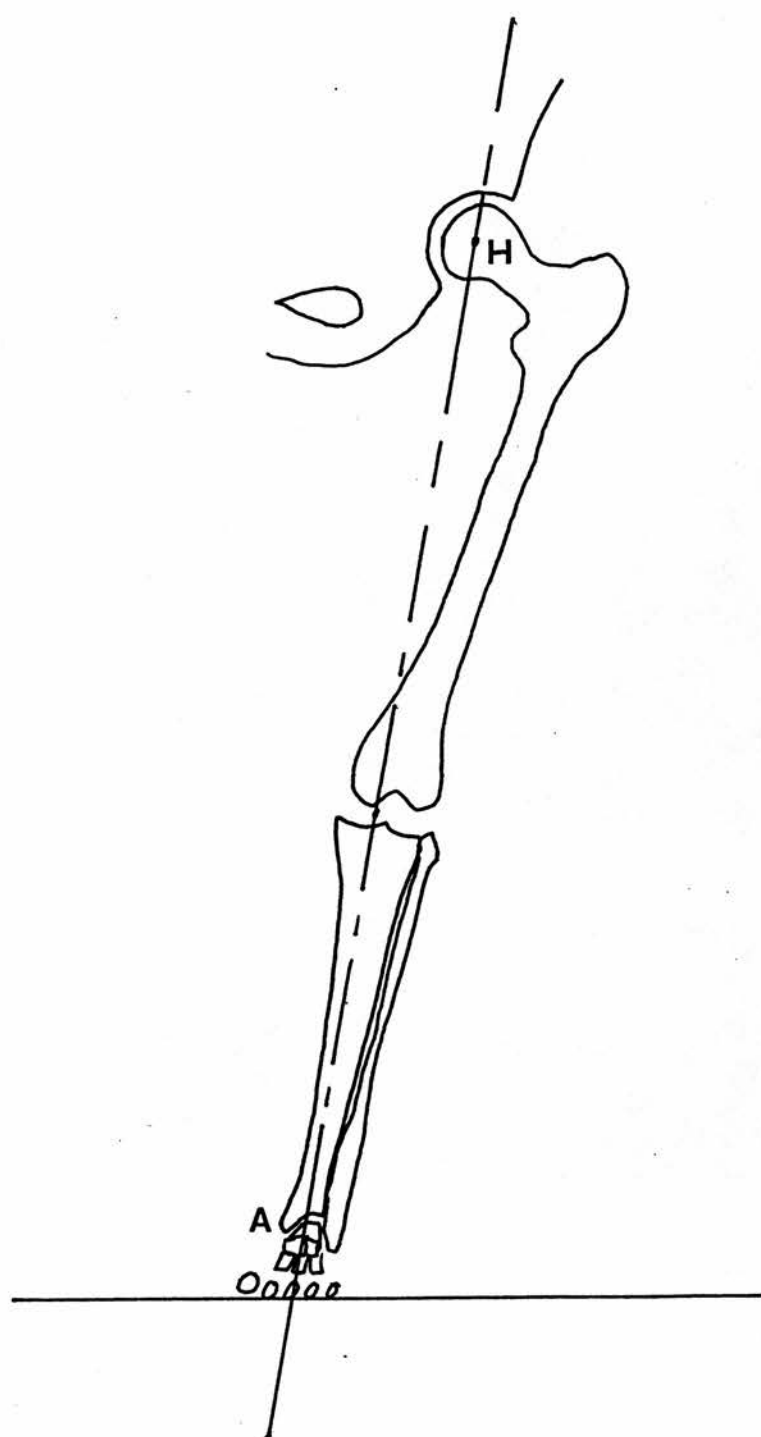


Fig. 3.5 Harrington's Y-axis of femoral axis system (Harrington 1974).

3.VII RESULTANT FORCES AND MOMENTS

Knowing the segment characteristics in terms of orientation to each other and to the ground, knowing the centres of rotation at the joints, and having information about the ground-foot reaction forces it is possible to calculate the intersegmental forces and moments at each joint by treating the limb under consideration as a free-body.

From the free-body diagram shown in Figure 3.6, the moments about the hip are given by :

$$\begin{aligned} \text{MHX} &= \text{FGZ} \cdot \text{HY} + \text{FGY} \cdot (\text{HZ} - \text{ZF}) \\ &= \text{FGZ} \cdot \text{HY} + \text{FGY} \cdot \text{HZ} - \text{MGX} \quad \text{_____} 3 \quad (15) \end{aligned}$$

$$\begin{aligned} \text{MHY} &= \text{FGX} \cdot \text{HY} - \text{FGY} \cdot (\text{HX} - \text{XF}) \\ &= \text{FGX} \cdot \text{HY} - \text{FGY} \cdot \text{HX} - \text{MGZ} \quad \text{_____} 3 \quad (16) \end{aligned}$$

$$\text{MHZ} = \text{FGZ} \cdot \text{HX} - \text{FGX} \cdot \text{HZ} + \text{MGY} \quad \text{_____} 3 \quad (17)$$

The knee moments are obtained by substituting the X,Y and Z coordinates for the knee joint centre.

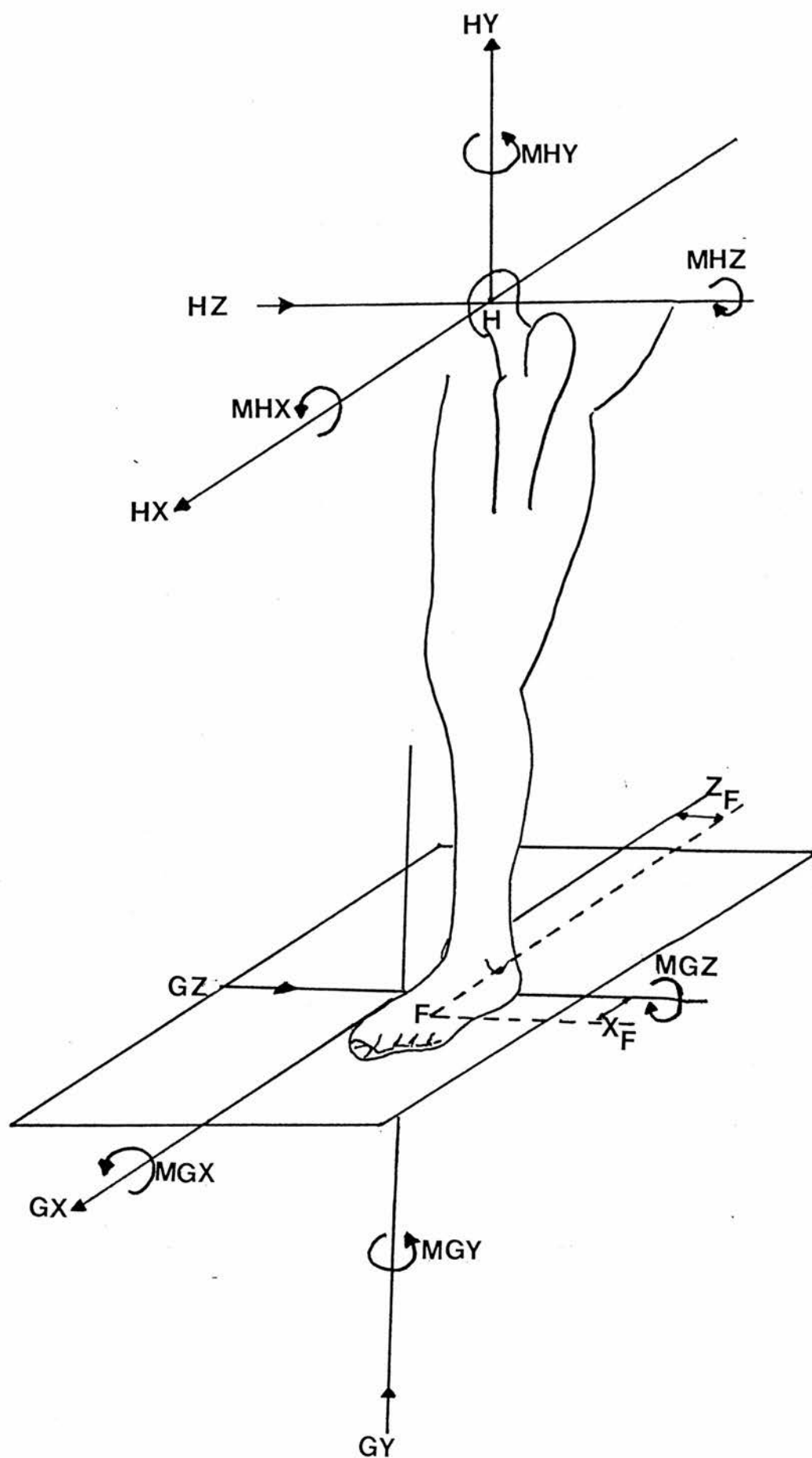


Fig. 3.6 Free body diagram.

3.VIII MUSCLE FORCES

The external force system acting as a joint is balanced by the forces acting between joint surfaces and tensioning of muscles and ligaments about the joint.

To calculate the muscle force four factors must be known :

1. The attachments of the muscles and their lines of action.
2. The relationship of the muscles to the joint axes.
3. The phasic activity of the muscles during the gait cycle.
4. The relative contribution to the balancing of the resultant moment that each muscle makes.

As has already been mentioned (1.IV 3.) the redundancy of the muscles about the hip and knee joints necessitate the formation of a model to enable the resultant forces and moments to be distributed amongst them. Whilst following the principles outlined above this model involves several simplifying assumptions. These are outlined below in the description of the model.

1. Muscle Model

- a) Muscle Grouping

In order to overcome the statically indeterminate force distribution problem it is necessary to reduce the number of unknowns to solve the equilibrium equations. This is achieved by reducing the number of muscles and ligaments considered at the joints. The muscles around the joints are grouped according to their primary function, and phasic activity during gait (Fig. 3.7).

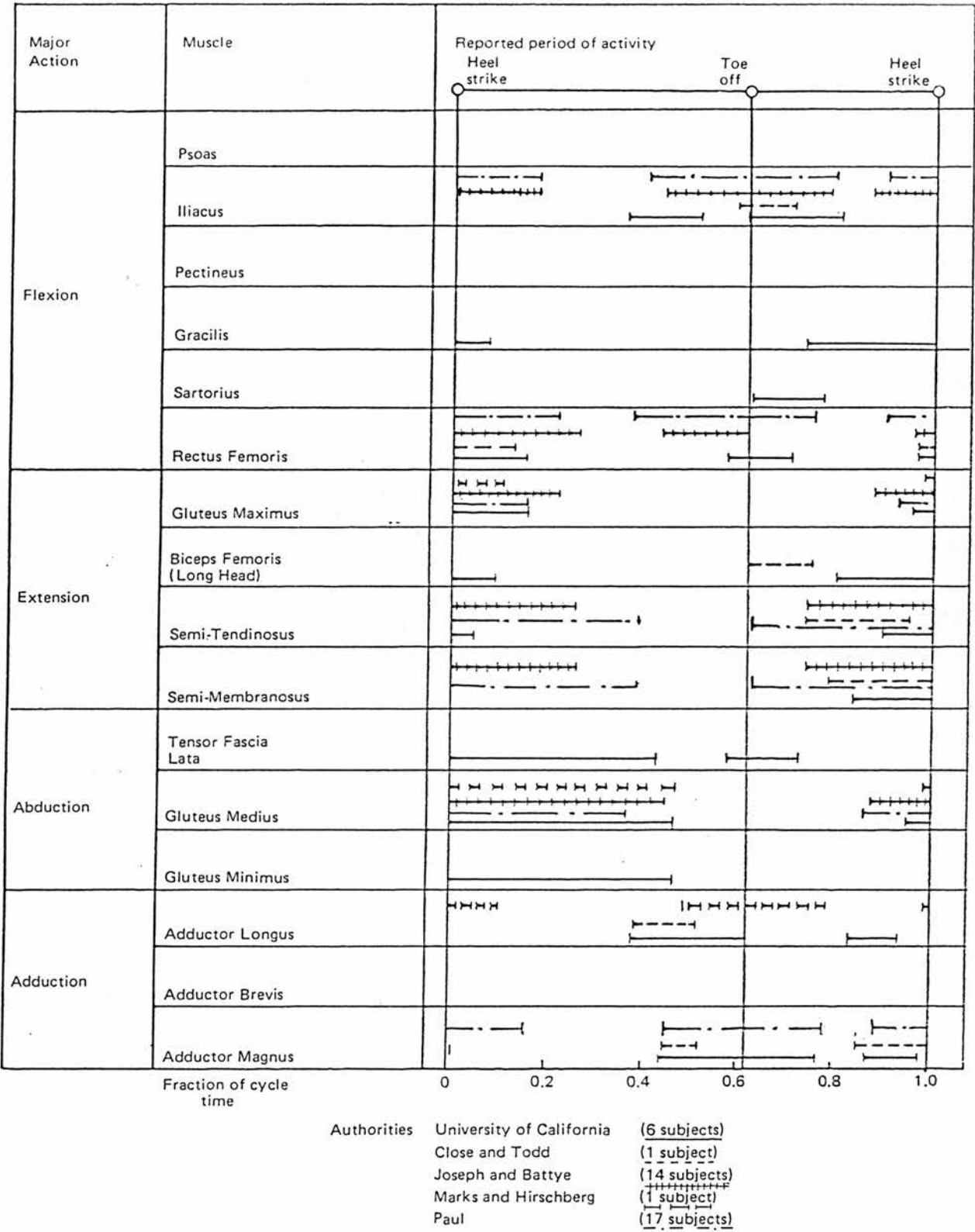
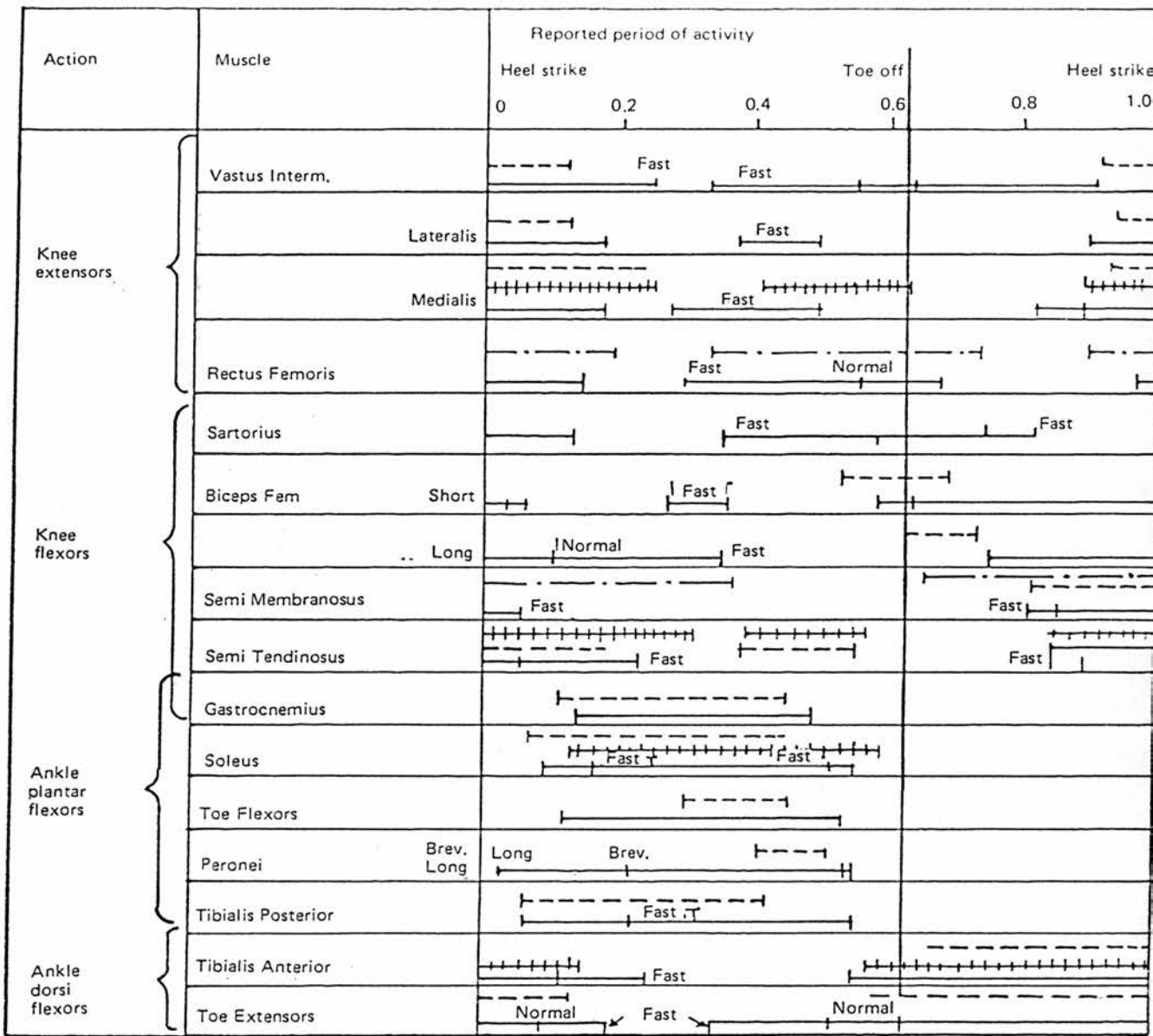


Fig. 3.7a Action and Phasing of muscles
 acting at the hip (Paul 1974).



Authorities	University of California	(6 subjects)
	Close and Todd	(1 subject)
	Joseph and Battye	(14 subjects)
	Paul	(17 subjects)

Fig. 3.7b Action and Phasing of Muscles acting at the knee and ankle (Paul 1974).

The small muscles around the hip - obturator internus and externus, the gemelli, piriformis and quadratus femoris, are deeply situated and little is known of their phasic activity. They are small in size and are rotators of the hip so that their contribution to loads across the hip joint is likely to be very small. They are therefore excluded from the analysis.

The remaining hip muscles are arranged into six groups after Paul (1968) (Fig. 3.8) :

1. Short flexors - psoas, iliacus
2. Long flexors - rectus femoris, sartorius
3. Short extensors - gluteus maximus
4. Long extensors - biceps femoris (long head)
semimembranosus,
semitendinosus
5. Abductors - tensor fascia lata, gluteus
medius, gluteus minimus
6. Adductors - adductor longus, brevis and
magnus, pectineus

At the knee joint the muscles were arranged into three groups :

1. Extensors - quadriceps femoris
2. Flexors - proximal - medial hamstrings
biceps femoris (long head)
semimembranosus
semitendinosus
3. Flexors - distal - soleus and gastrocnemius

This system of functional muscle grouping has been used in previous studies reported from the Bioengineering Unit at the University of Strathclyde, and is justifiable according to the work of Marks and Hirschberg (1958) and the University of California (1953).

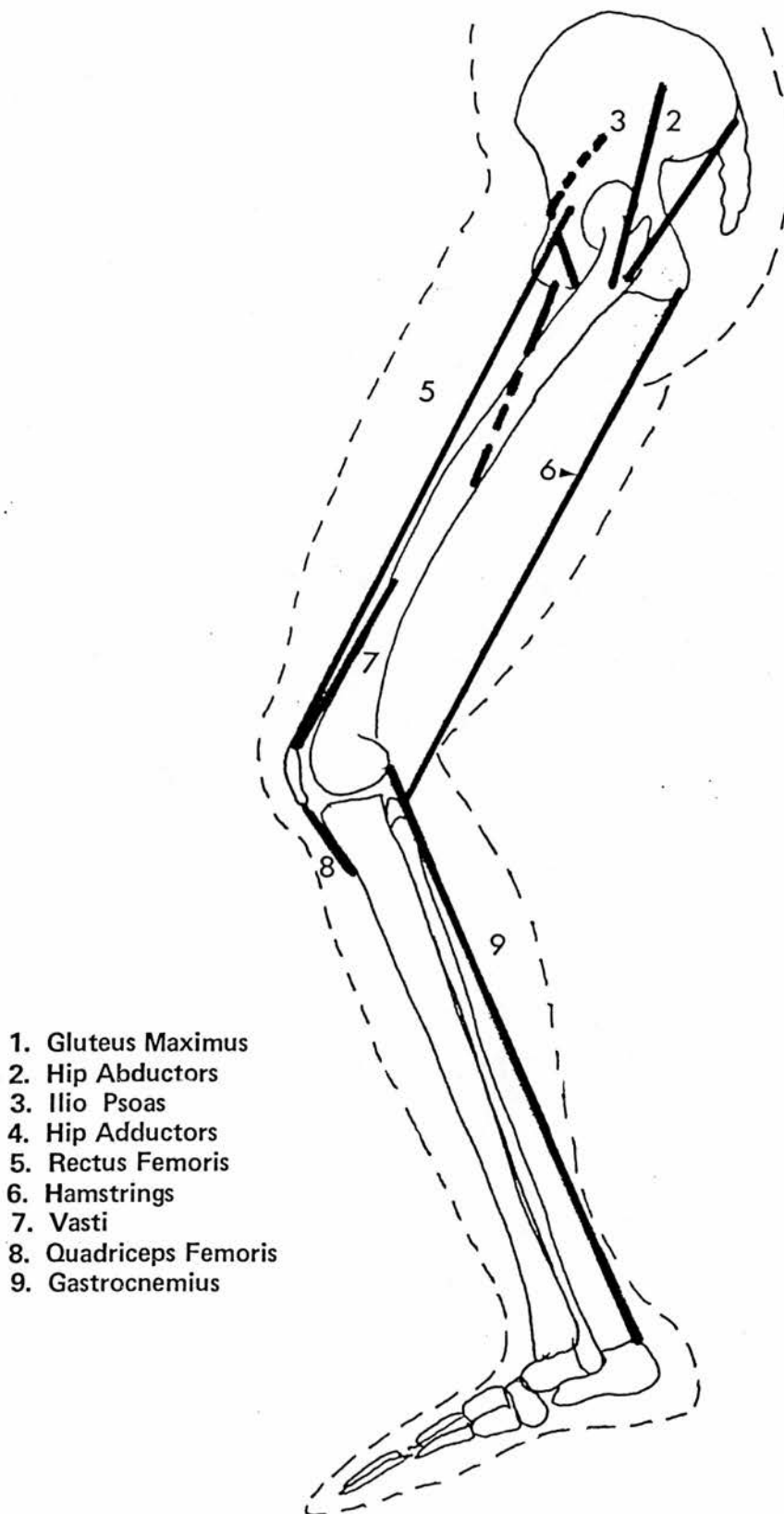


Fig. 3.8 Lines of action of muscle groups at hip and knee (Tooth 1976).

b) Muscle attachments and lines of action

The origins and insertions of the muscles under consideration were taken from the anthropometric data of Morrison (1967) and Poulson (1973). At the hip the centroid of the group attachment was given coordinates based upon the centre of rotation of the joint as origin for both femoral and pelvic attachments. The muscles about the knee joint were treated similarly and given coordinates about the knee joint centre in the tibial axis reference frame. Scaling factors were used to relate the dimensions of the dissected cadavers to the patient under test.

In the original analyses using this system (Poulson, 1973, McGrouther, 1974, Tooth, 1976) the position of the hip joint centre was calculated from the distance between the anterior superior iliac spines (CB, distance a) using a system of linear three-dimensional proportionality based upon the cadaver dissections. Since the present study is specifically investigating the effects of anatomy at the hip joint upon gait, changes had to be introduced at this stage.

The mean of the coordinates of the healthy, contralateral hip joint centre, based upon the anterior superior iliac spine as origin, corrected for magnification differences and standardised for pelvic dimensions according to the distance CB, was estimated from the radiographic measurements and used as the standard hip joint centre position (Fig. 3.4). Its coordinates in the femoral axis system were determined in the same way.

The differences in location of the joint centre of the diseased or replaced hip from the standard hip were then inserted into the data file for both the pelvic and femoral axis systems to define the centre of rotation of the affected joint. The origins and insertions of the muscles are determined with reference to the hip joint centre predicted by proportionality, therefore inserting new coordinates for the hip joint centre does not affect the skeletal coordinates for the muscles. Hence, the muscle to hip joint centre relationships approximate to the pathological situation, bearing in mind the fact that the initial scaling does not take account of the differences in shape of the appropriate structures between the cadaver and patient under test, and no allowance can be made for the effect of a change in a muscle's length upon its function when using this muscle model.

The line of action of any given muscle group was obtained by drawing a straight line from the centroid of attachment on one skeletal structure to the centroid of attachment on the other. Such a line is obviously an approximation, since, because of their own shape, and the volume, and shape of the underlying muscles the true muscle centre lines will be curved. However, the effect of this on lever arms relative to joint axes will be small since the distances between the joint axes and areas of insertion of the tendon is small.

It is not always possible to connect the muscle origin and insertion using a straight line. This is sometimes the case during the gait cycle for quadriceps

femoris, gluteus maximus and ilio-psoas. The quadriceps femoris acts upon the shank through the patellar tendon and its line of action was taken as being coincident with this tendon. Morrison (1967) showed that this line of action was dependent upon the knee flexion angle and it can be calculated using the mathematical relationships derived by Morrison.

Paul (1967) found that certain hip angles gluteus maximus was deflected around the ischium (Fig. 3.9a) and he devised the criterion that if the distance between the insertion on the pelvis and insertion on the femur exceeded a limiting value (the distance between insertion and origin when the line joining them was a tangent to the ischium) a new pelvic origin would be assigned.

Paul also found that the line of action of iliopsoas was not a straight line (Fig. 3.9b) when the angle between the pelvis and femur was greater than a certain value. He therefore gave it a radius in the XY plane and calculated the Z coordinates for the normal origin and insertion.

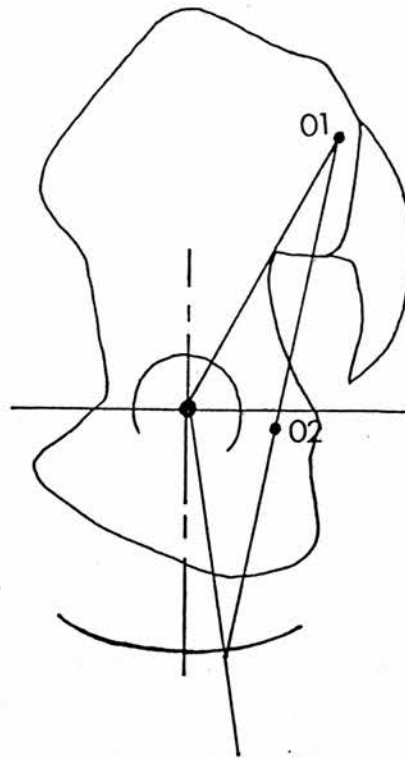
c) Calculation of Muscle Forces

Several further assumptions have to be made in order to use the model outlined above.

1. The direction of the resultant force of a muscle group remains constant regardless of the intensity of the contraction. Allowance cannot be made for the differing force actions in different muscle fibres which determine the line of action of the resultant.

2. The analysis is aimed at determining the probable maximum and minimum values of joint force and the principle of minimal muscle force is observed.

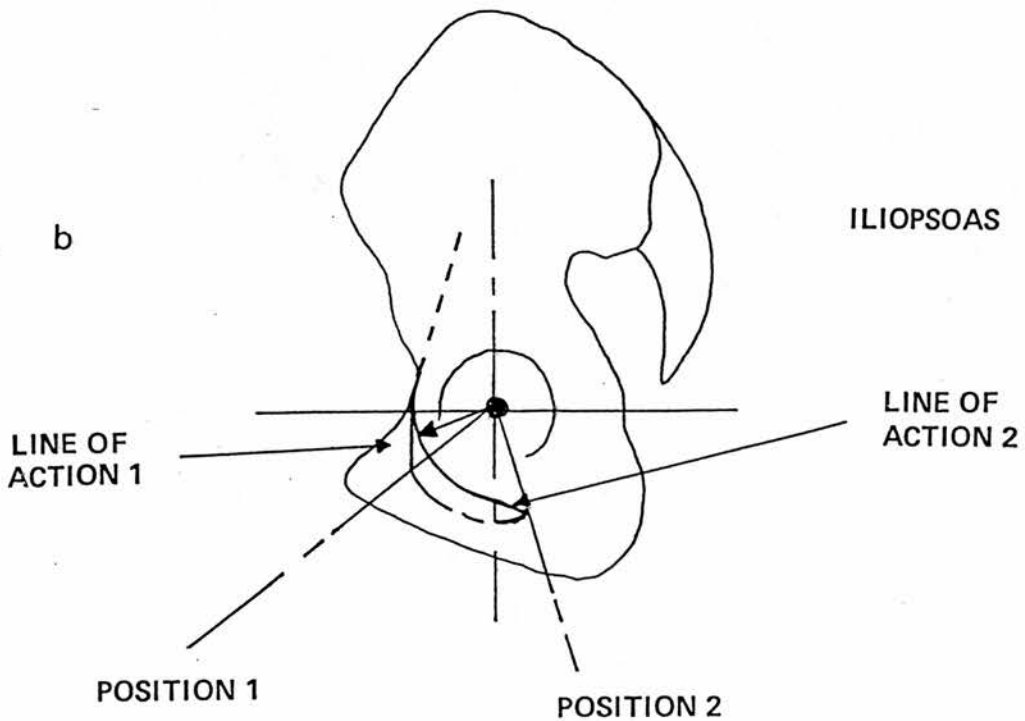
a



GLUTEUS MAXIMUS

Fig. 3.9a Change in line of action of gluteus maximus with hip flexion
 01 - origin 1
 02 - origin 2
 (Paul 1967)

b



ILIOPSOAS

Fig. 3.9b Change in line of action of iliopsoas with hip flexion
 (Paul 1967).

3. No antagonistic activity takes place between opposing muscle groups. This is clearly not the case as an examination of Figure 3.7 will show. However, as mentioned in Chapter 1.IV the factors determining antagonistic activity are unknown and therefore cannot be allowed for. The effect of this assumption will be to slightly underestimate the joint force at certain phases of the cycle, although probably not at periods of maximum load.

Even with the major simplifying assumptions with regard to muscle anatomy it can be seen from Figure 3.8 that there is still a situation of statical indeterminacy, there being 2-joint muscles at the hip and knee - hamstrings, tensor fascia lata and rectus femoris (quadriceps femoris). In this analysis the choice of active muscle groups is based upon the resultant moment about the hip joint under consideration. If there are resultant moments at the hip and knee in the same direction one or other of these 2-joint muscles will be advantageously placed to transmit the force. In this situation there are three possibilities :

(1) The relevant 2-joint muscle could balance the moment that requires least force from it leaving the residual moment to be balanced by the appropriate one joint muscle.

(2) The tension in the 2-joint muscle could equilibrate the moment at the hip joint requiring the greater tension, the reversal of the moment at the other joint being balanced by the one joint antagonist.

(3) A situation between the limits of (1) and (2) could exist.

Without using some form of complex optimisation technique (1.IV) option (3) cannot be coped with. Option (2) would involve increasing the moment at the other joint requiring greater energy expenditure in the form of antagonistic activity, and increased joint load. It is therefore thought to be physiologically unlikely and option (1) will be used in the analysis.

d) Knee Musculature

At the knee joint all three muscle groups are considered to act about the medio-lateral axis (Z-axis) of the tibia.

If the resultant moment about this axis (MKZ) is negative the quadriceps femoris group was assumed to be active. A positive value for MKZ means that either hamstrings or gastrocnemius could be acting. The resolution of this dilemma is dependent upon the sign of the moment about the Z-axis at the hip joint (MHX). If this moment was negative, tending to extend the thigh on the hip, activity in the hamstrings would increase the moment at the hip. Therefore in this situation gastrocnemius would be selected as being the only active muscle group at the knee.

If MHZ was positive then either the force required in the hamstrings to balance the hip moment may be less than that required to balance the knee moment or the converse. By applying option (1) above hamstring force would be evaluated from MHX and the residual moment required at the knee would be provided by gastrocnemius in the former case, and in the latter case the hamstring

force would be determined from the knee moment, the residual balancing moment at the hip being provided by the hip musculature.

e) Hip Musculature

Since the muscles acting about the Y-axis (the rotators) have not been considered in the model the equilibrium equations about this axis are not used.

The tension in the hamstrings is known from the analysis at the knee (vide supra) and the tension in rectus femoris cannot exceed that in quadriceps femoris, which was also found at the knee. Despite this, the situation remains indeterminate, there being two equilibrium equations and five unknowns.

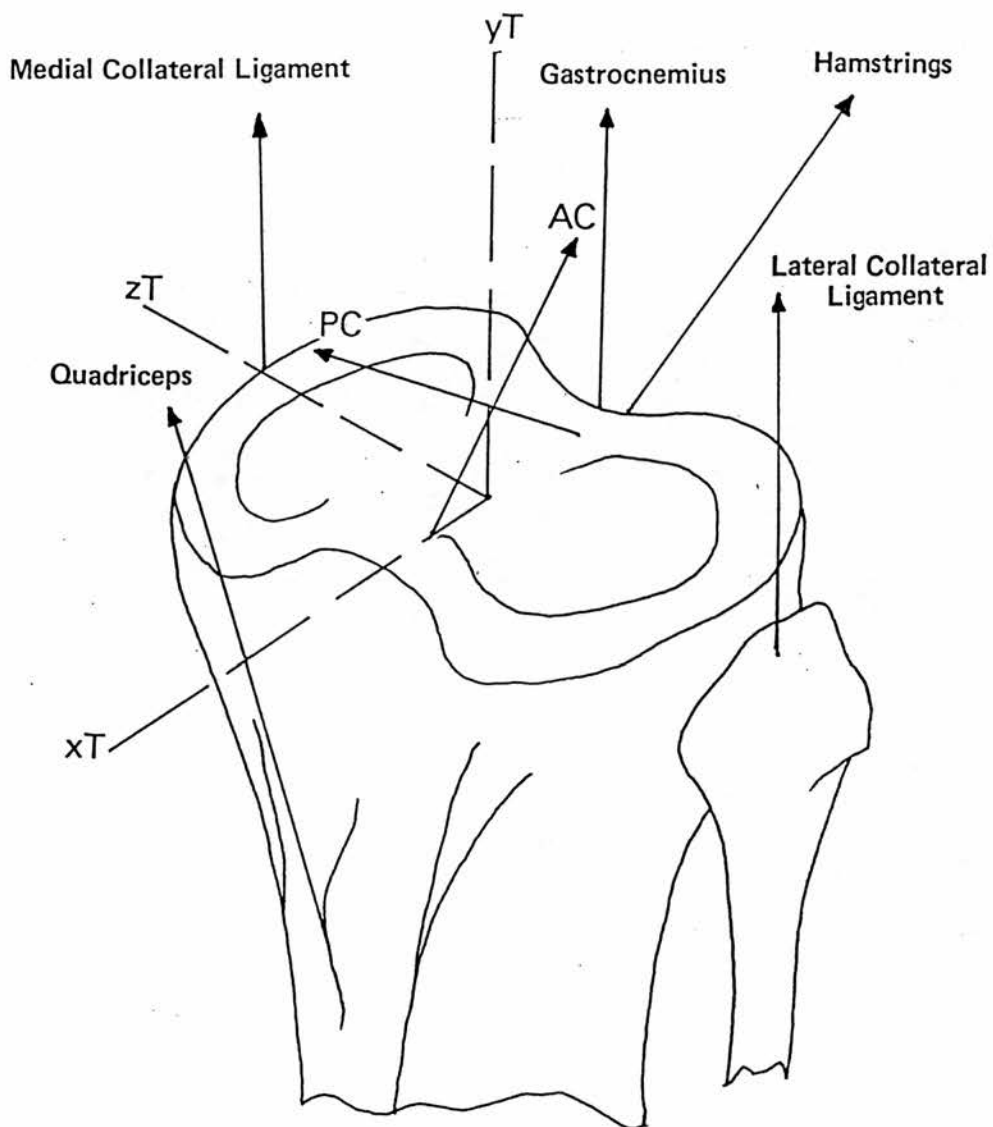
If one assumes three of the groups to be inactive it is possible to calculate the tension in the remaining two groups. As in the elimination procedure of Chao, Opgrande and Axmear (1976) (1.IV 3) the equations can be solved for all possible pairings of the muscle groups, allowing for any hamstrings activity. Only certain solutions will be acceptable and the criteria are :

(1) Only those solutions with both values positive are chosen since muscles can only act in tension.

(2) If more than one pairing meets condition (1) the pair whose sum is the least is accepted.

(3) When rectus femoris is acting its tension cannot be greater than that found in quadriceps femoris. If it is, the solution is rejected and rectus

femoris takes the value of the tension in quadriceps femoris. The hip moments are again adjusted allowing for the contribution of rectus femoris and the equations are solved considering only gluteus maximum, abductors, adductors and iliopsoas to be active.



AC – ANTERIOR CRUCIATE LIGAMENT

PC – POSTERIOR CRUCIATE LIGAMENT

Fig. 3.10 Muscle and Ligament Force
System at the Knee.

IX JOINT FORCES

1. Knee Joint Force

All forces at the knee joint are assumed to be transmitted by the muscles, collateral ligaments, and cruciate ligaments (Fig. 3.10). All the muscle forces can be determined as above but in order to calculate the resultant joint load the forces in the ligaments must be calculated. The role of the posterior capsule in resisting moments about the tibial X-axis when the joint is near its fully extended position (only twice in a normal knee movement pattern) cannot be determined and is ignored. This will result in a slight under-estimation of joint load.

a) Cruciate Ligaments

The cruciate ligaments are assumed to resist only antero-posterior forces (Morrison, 1967, Harrington, 1974 and 1983) and tension in one ligament implies no tension in the other. Their possible role in resisting torsional effects is ignored since moments about the Y-axis are ignored.

b) Collateral Ligaments

A moment about the X-axis of the knee tending to abduct the shank on the femur will result in a shift of the centre of pressure at the knee towards the lateral side of the joint (Fig. 3.11). As the centre of pressure moves laterally a greater proportion of the total load will be taken by the lateral joint compartment. Once the medial compartment load has reached zero any further abduction moment will be balanced by tension in the medial collateral ligament. The converse is true for an X moment

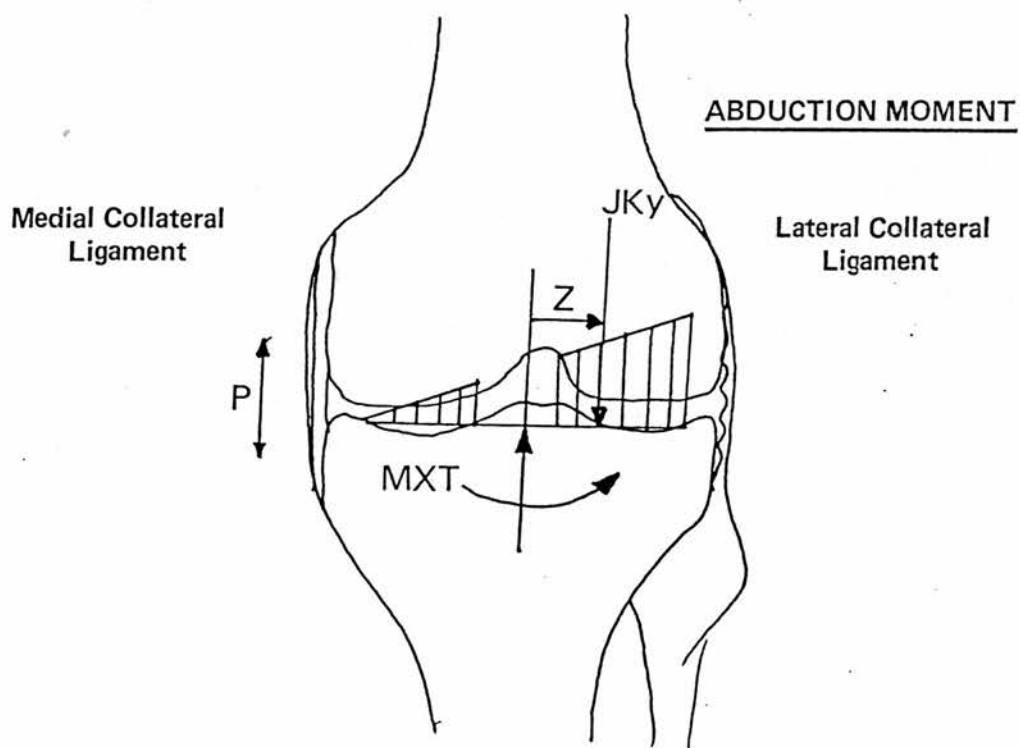
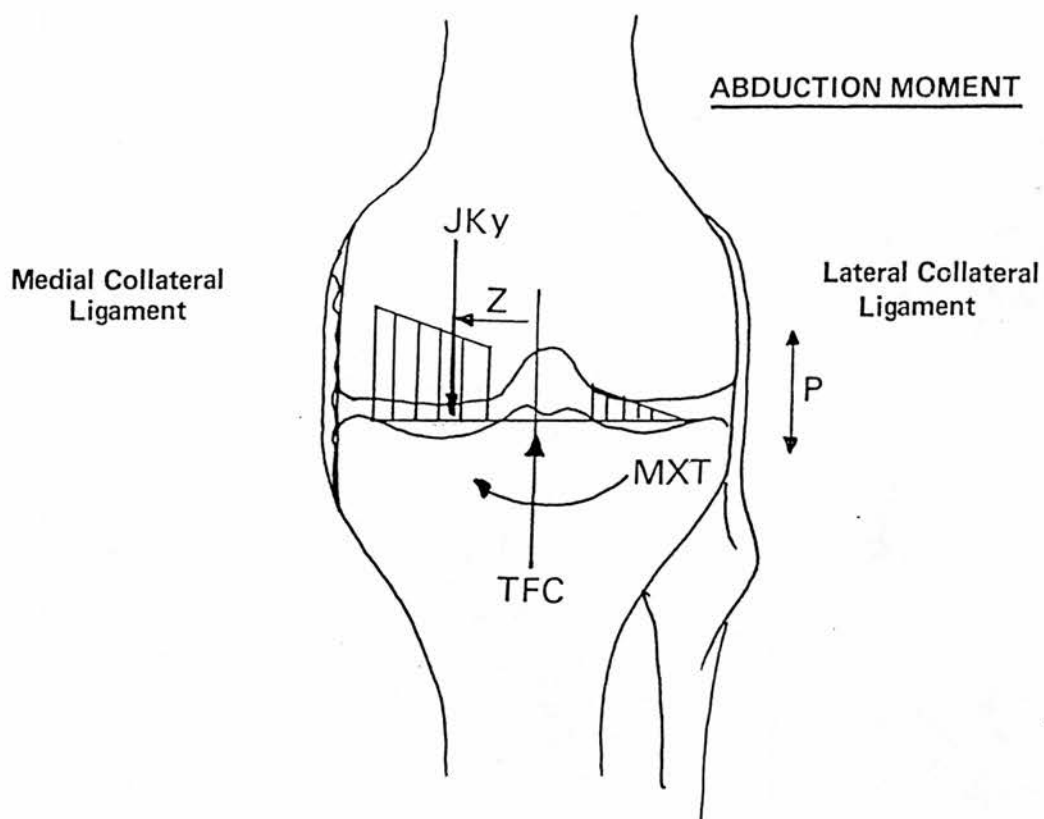


Fig. 3.11 Effect of abduction/adduction moments about the knee joint upon compartmental load and ligamentous tension (Morrison 1967).

at the knee of opposite sign, in which case tension is taken to exist in the lateral collateral ligament. The positions of the centre of pressure at which tension occurs in the collateral ligaments is known as a limiting value.

The limiting value of the position of the centre of pressure is scaled to the patient under test from the measurements made on cadaveric knees by Morrison (1967) using measurements made at the time of testing. Morrison took the position of the limiting value in the ZY plane to be the Z value of the centre of the medial compartment.

The need to invoke tension in the ligament can then be determined by considering the sign of the moments about the X-axis. If tension is present in a ligament the solution for joint load is calculated for that ligament and the loaded contralateral compartment. If no tension is present in the ligaments the joint load is calculated for the two compartments.

2. Hip Joint Force

The muscle forces in all six groups of muscles about the hip are known from the foregoing analysis. These muscle forces are divided into their components along the coordinate axes at the hip and the resultant hip joint force is obtained by vector addition of these components.

CHAPTER 4

RESULTS

I INTRODUCTION

II HIP DISEASE GROUP

1. Patient Test

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the
Hip and Knee
- e. Muscle Forces
- f. Joint Loads

2. Summary of Results for Hip Disease Group

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the
Hip and Knee
- e. Muscle Forces
- f. Joint Loads

III HIP JOINT REPLACEMENT GROUP

1. Patient Test

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the
Hip and Knee
- e. Muscle Forces
- f. Joint Loads

2. Summary of Results for Hip Joint Replacement Group

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the
Hip and Knee
- e. Joint Loads
- f. Phasing of Hip Joint
Force Maxima.

IV GIRDLESTONE EXCISION ARTHROPLASTY GROUP

1. Results

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the
Hip and Knee
- e. Muscle and Joint Forces

V THE EFFECT OF ANATOMICAL VARIABLES ON
JOINT FUNCTION

1. Anatomical Variables

- a. The Degenerative Hip Joint
- b. The Replaced Hip Joint

2. Statistical Analysis

- a. Multiple Linear Regression
Analysis
- b. Correlations

VI SUMMARY OF RESULTS

- 1. Hip Disease Group
- 2. Hip Joint Replacement Group
- 3. Girdlestone Excision Arthroplasty
Group
- 4. Anatomical Variables and
Hip Joint Function

4.I INTRODUCTION

The results will be set out in two parts. In the first part the results from the gait studies in the two groups of patients will be presented.

The composition of each group was described in Chapter 2. The pre-operative results from the Hip Joint Replacement Group (The Hip Disease Group) will be considered to permit the study of the effect of hip disease upon gait. The variations between this group and normal subjects (Paul, 1967) will be discussed with reference to the results of one patient. The results for the entire group will then be summarised (Chapter 4.II).

The findings in the post-operative Hip Joint Replacement Group will be presented, attention again being paid to the results of one patient (Chapter 4.III.1). These results will be related to the pre-operative results and to the results seen in normal subjects. Subsequently, the results for the entire group will be summarised (Chapter 4.III.2) and the data for the Excision Arthroplasty Group will be presented (Chapter 4.IV) and considered in relation to the findings in the pre-operative and Total Hip Replacement Groups.

In the second part of this Chapter, the influence of the anatomical variables at the hip joint upon the joint loads will be considered for the Hip Disease and Hip Joint Replacement Groups (Chapter 4.V).

With the exception of the time-distance parameters of gait all other results are presented as graphs of the

variation in the parameter with the duration of stance phase. To facilitate the comparison of results from different patients and between the limbs of the same patient, the stance phase duration has been normalised. For the same reason all forces are expressed in multiples of body weight.

CHAPTER 4 TABLE 1. CLINICAL DETAILS HIP DISEASE GROUP PRE- AND POST-OPERATIVELY

Patient	Time Opn.	HARRIS SCORE		RANGE OF MOVEMENT (degrees)					Leg Length Discre- pancy cm.	Trendelenburg Test	Total Harris Score
1	Pre	20	23	15- 90	0	0	20-40	-20	0.0	-	44
	Post	44	47	0-100	0-20	0-20	10-40	-10	0.0	-	95
2	Pre	10	20	45- 85	-10	10	15	-15	3.0	+	31
	Post	44	35	0- 95	0-20	0-20	0-20	0-10	2.0	-	81
3	Pre	20	36	5-120	-15	15-30	0	0-10	1.0	±	60
	Post	44	47	0-120	0-30	0-30	0-15	0-15	1.0	-	96
4	Pre	20	30	10-100	0-25	0-20	0-20	0-15	0.0	-	52
	Post	44	47	0- 90	0-30	0-30	0-30	0	2.0	-	93
5	Pre	20	18	0- 90	0-15	0-10	0-15	0-10	0.0	+	42
	Post	44	47	0-100	0-20	0-15	0-15	0-15	0.0	-	96
6	Pre	40	26	25- 90	0-15	0-15	10-30	-10	2.5	+	70
	Post	44	47	0-100	0-30	0-20	0-20	0	0.0	-	93
7	Pre	20	34	0-100	0-30	0-20	0-30	0-10	1.0	-	57
	Post	40	47	0- 90	0-30	0-30	0-30	0-20	1.5	-	94
8	Pre	20	28	20- 90	0-20	0-10	0-20	0	1.5	±	53
	Post	44	43	0- 90	0-20	0-20	0-30	0-10	0.0	-	93
9	Pre	20	31	10- 90	0- 5	0- 5	0-10	0-15	1.5	+	52
	Post	44	40	0-100	0-30	0-25	0-15	0-10	1.0	±	90

NOTE: Patient 4 showed 10 degrees of hip extension.
No hip extension was measurable in any other patient.

CHAPTER 4 TABLE 1 continued. CLINICAL DETAILS HIP DISEASE GROUP PRE- AND POST-OPERATIVELY

Patient	Time Opn.	HARRIS SCORE Pain Function		RANGE OF MOVEMENT (degrees)						Leg Length Discrepancy cm.	Trendelenburg Test	Total Harris Score
10	Pre	20	23	15- 70	0-15	0- 5	20-30	-20	2.0	+	45	
	Post	44	44	0-110	0-30	0-30	0-30	0-20	0.0	-	89	
11	Pre	10	19	10-100	0-10	0-10	0-10	0	0.0	+	33	
	Post	-	-	-	-	-	-	-	-	-	-	
12	Pre	20	27	20- 80	0-25	0-10	0-10	0	0.0	+	49	
	Post	44	47	0- 90	0-25	0-20	0-30	0-10	0.0	-	94	
13	Pre	20	35	15-110	0-10	0-20	0-10	0-10	0.0	+	57	
	Post	44	47	0-100	0-20	0-20	0-20	0-20	0.0	-	93	
14	Pre	10	22	15- 90	-10	10-20	10-20	-10	2.0	+	34	
	Post	44	43	0- 90	0-30	0-20	10-30	0	0.0	-	90	
15	Pre	10	24	10- 90	0	0	20	-20	1.0	+	36	
	Post	44	43	10-100	0-10	0-15	0-15	0	0.0	-	91	
16	Pre	20	28	0- 90	0- 5	0- 5	0-30	0-10	2.5	-	51	
	Post	44	41	10-100	0-30	0-20	0-10	0-10	2.0	-	87	
25	Pre	-	-	-	-	-	-	-	-	-	-	
	Post	40	33	0- 90	0-30	0-30	0-40	0	2.5	+	76	

NOTE: Patient 14 showed 10 degrees of hip extension.
No hip extension was measurable in any other patient.

4.II HIP DISEASE GROUP

This Group comprised 15 patients, two of the Total Hip Replacement Pre-Operative Group (patients 4 and 7) being omitted because of the presence of femoral head replacements. The clinical details of these patients can be found in Table 4.1.

1. Patient Test

The peak values for the results in all of these patients will be found in Table 4.2 at the end of Section 4.III. The results for patient 15 will now be considered in detail under the headings:

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the Hip and Knee
- e. Muscle Forces
- f. Joint Loads

Patient 15 was a 59-year-old woman with a three-year history of a painful right hip which was the site of osteoarthritis. She was 1.63 metres in height and weighed 66.9 kg.

a. Time-Distance Parameters

Stride-length and width, and cadence were not measured. The mean walking velocity was 0.84m.s^{-1} (normal 1.40m.s^{-1}). The duration of stance phase on the affected right lower limb was 0.76 sec. and 0.82 sec. on the healthy limb.

AFFECTED LIMB

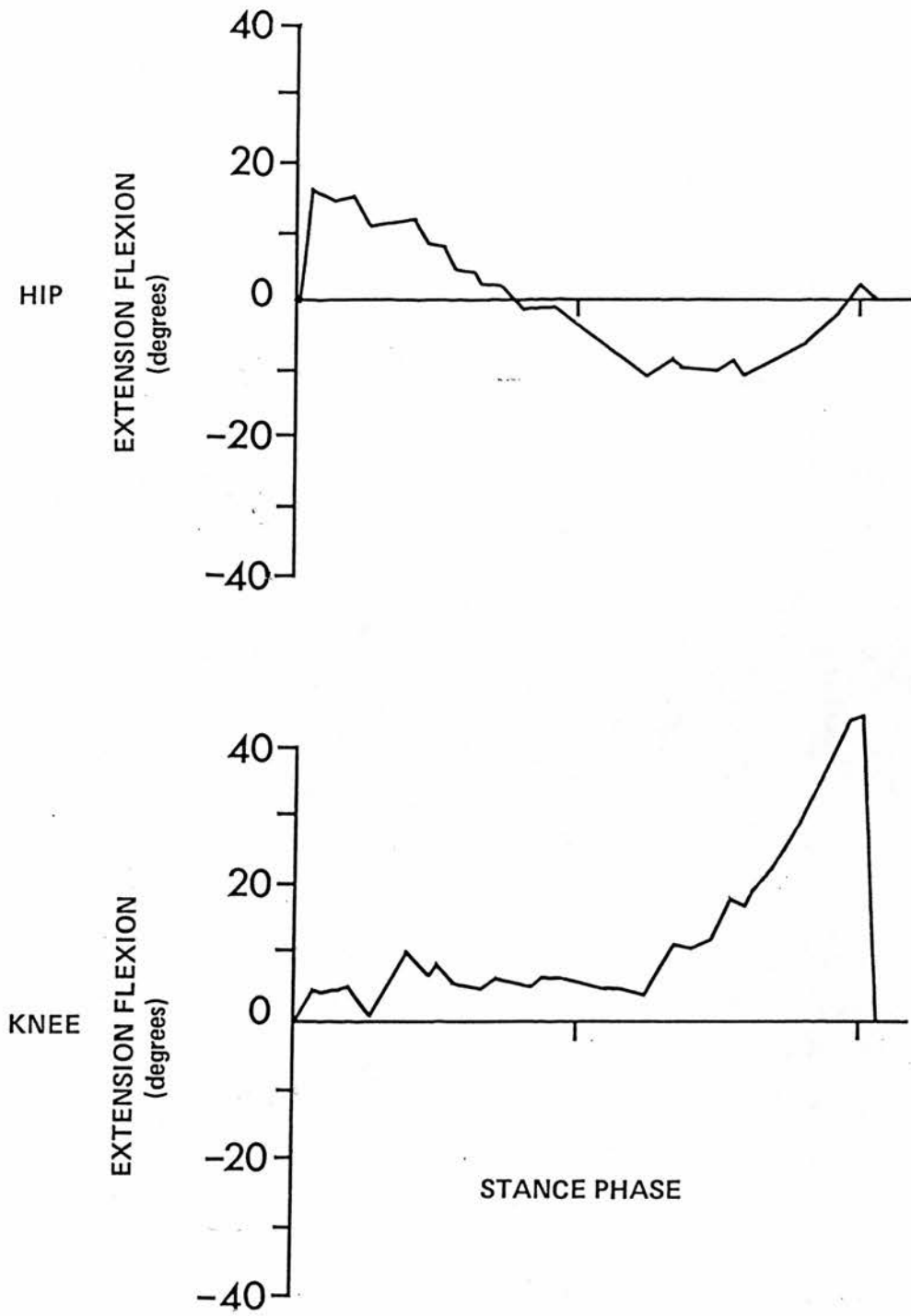


Fig. 4.1a Range of movement at hip and knee of affected limb.

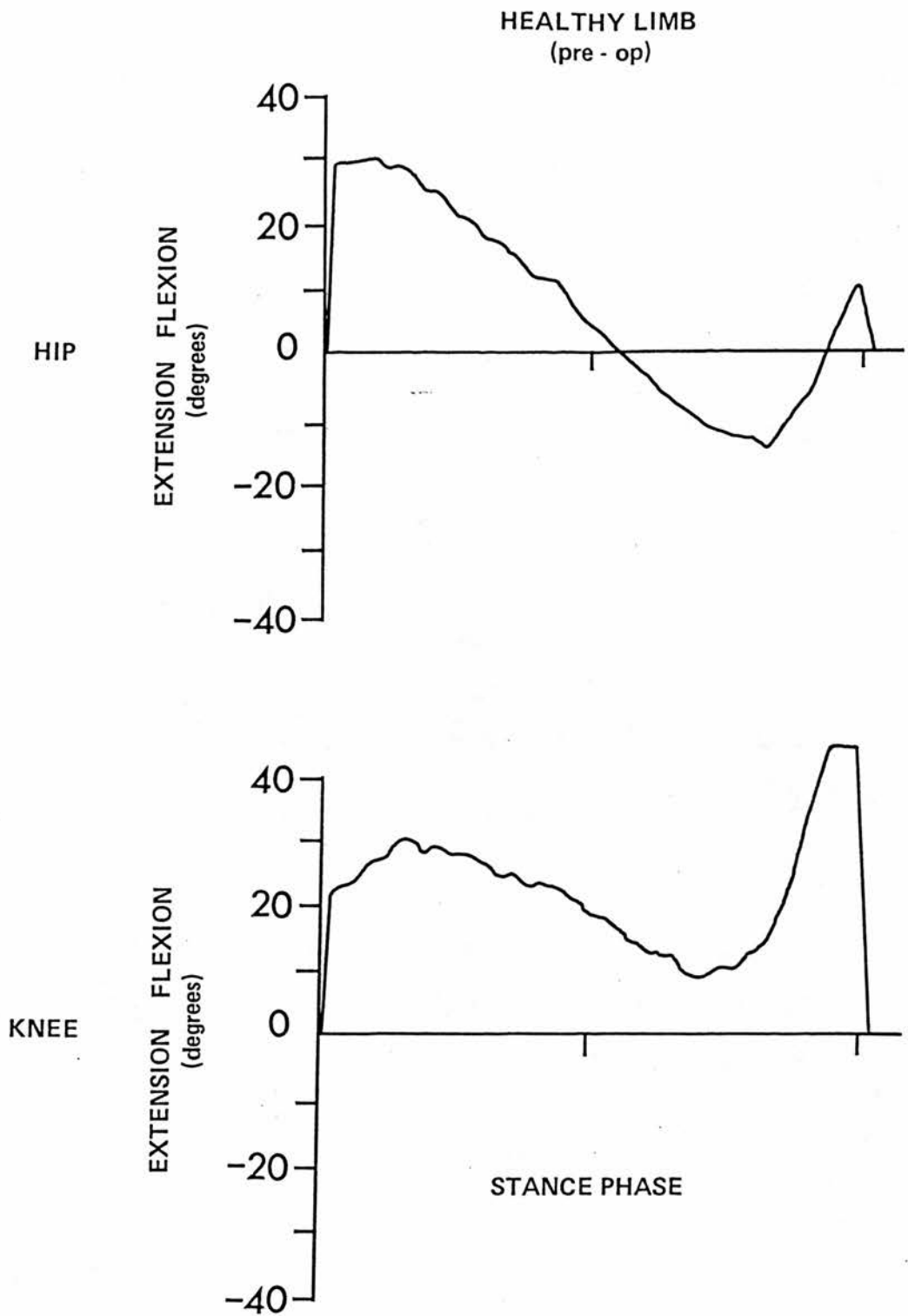


Fig. 4.1b Range of movement at hip and knee of healthy limb.

b. Limb Movements

Figures 4.1a and b represent the mean movement traces at the hip and knee joints for two test walks on each limb. There is a striking asymmetry of movement at both joints.

Hip Joint

At the affected hip there is from 17° flexion to 11° extension as compared with 30° of flexion and 14° of extension on the healthy side. Also the change from flexion to extension occurs at 40% of stance phase in the affected limb compared with 56% and 67% in the healthy limb. In the normal subjects 23° of flexion and 21° of extension with a changeover at 50% of stance is seen.

Knee Joint

At the knee joint of the affected side the usual triphasic pattern (Morrison, 1967, Chapter 1) is lost, the knee remaining near full extension until beyond 60% of stance when it begins to flex and continues to do so until toe-off. On the healthy side the triphasic pattern remains but the knee flexes excessively to 30° (normal 22°) and fails to go into as much extension as the normal knee (2°) at mid-stance.

c. Ground-Foot Reaction Forces

Both limbs show the initial sharp peak in FXG not seen in all patients, which results from the heel striking the ground sharply (Figs. 4.2a and b). There is then a force acting against the direction of progression until midstance when there is an increasing force in the direction of progression until toe-off. The magnitude

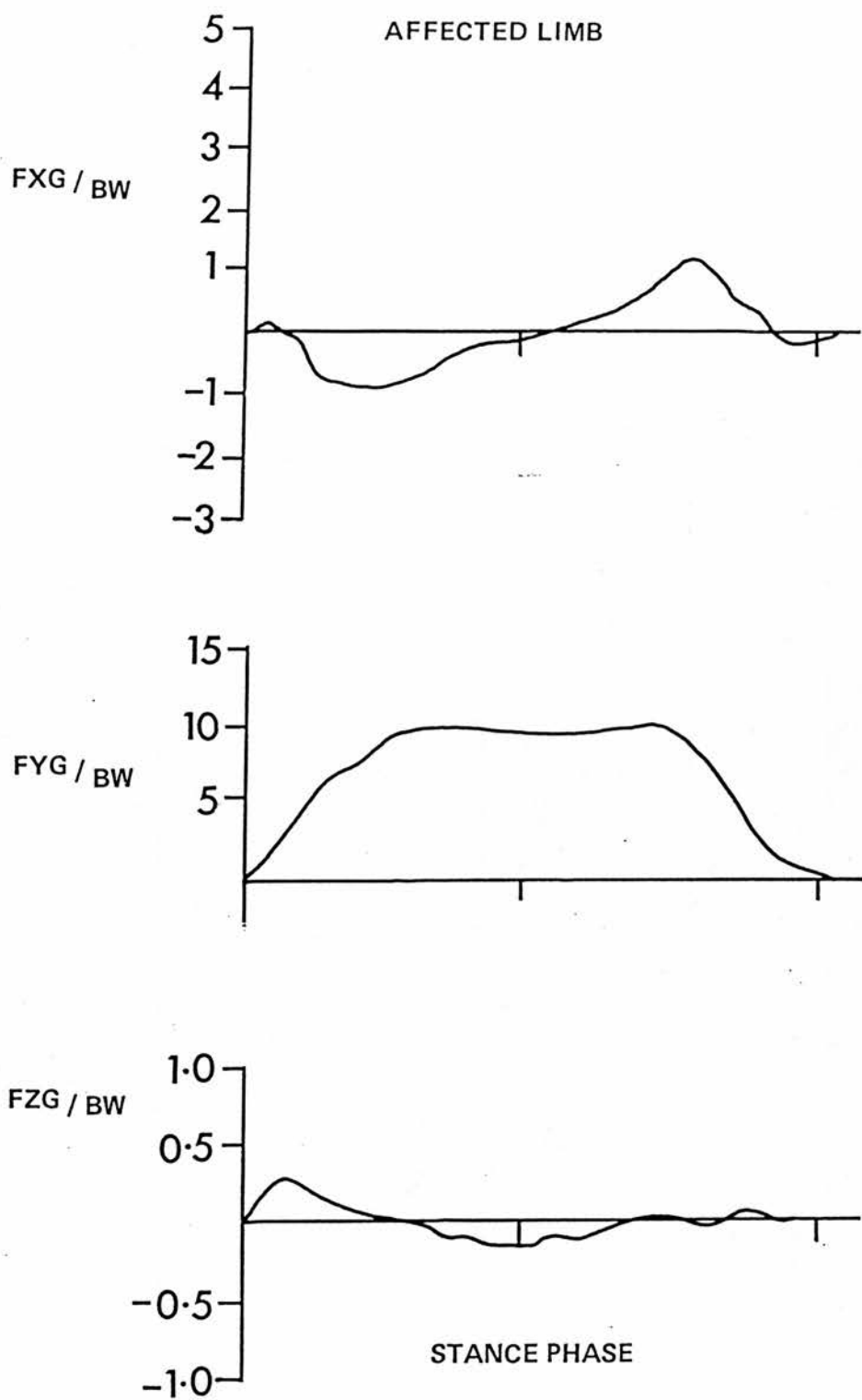


Fig. 4.2a Ground reaction forces to affected limb

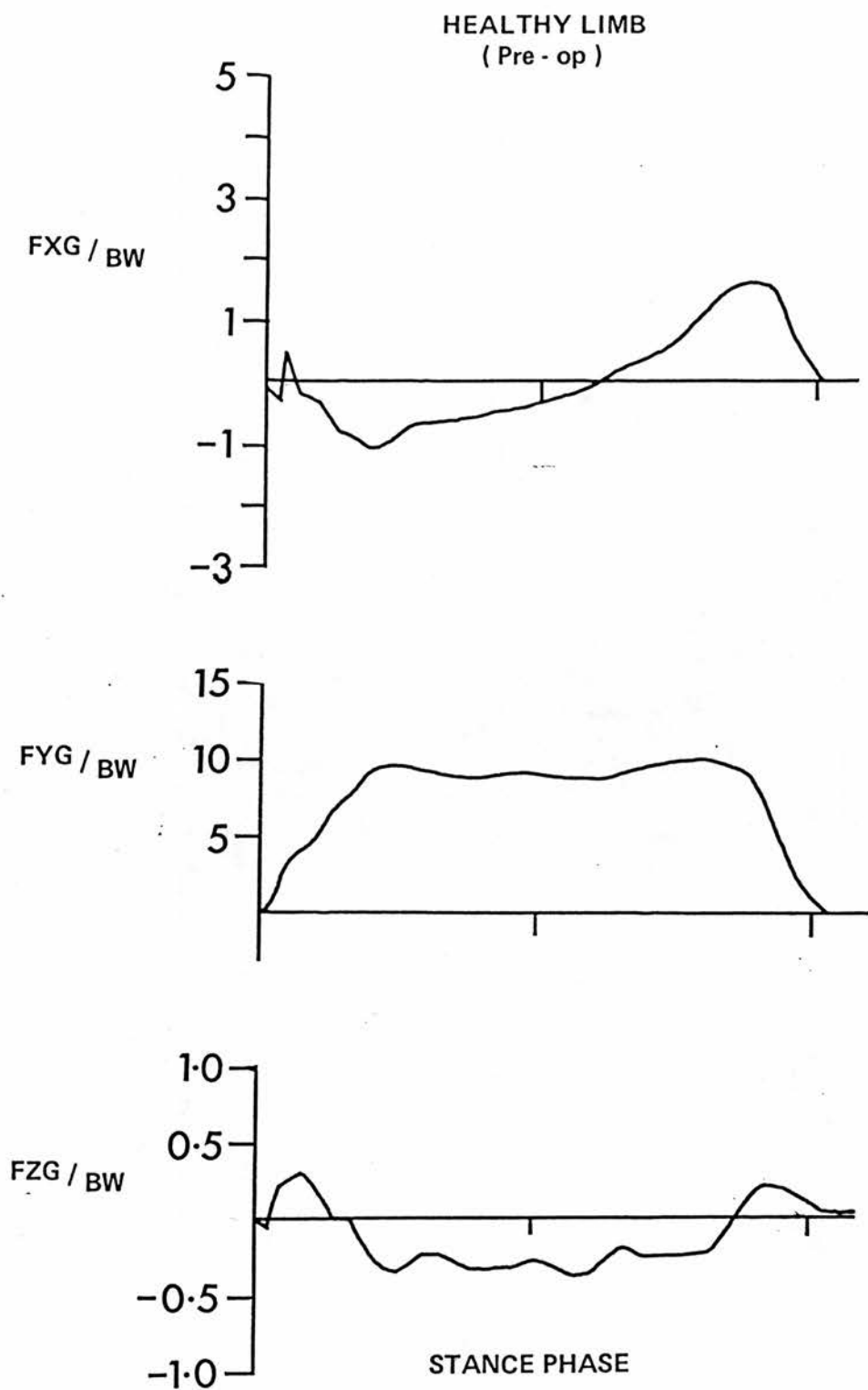


Fig. 4.2b Ground Reaction Forces to Healthy Limb.

of both the fore and aft forces is greater on the healthy side, but less than seen in the normal subjects.

Neither limb exhibits the typical two peak and trough pattern of the vertical component of force, FYG, and the timing of the post heel-strike peak of load in the affected limb is later than that in the healthy limb indicating a slower rate of loading. The magnitude of the two curves is approximately the same.

FZG represents the shearing force between the foot and the floor in the medio-lateral direction. The variability of this seen in Figure 4.2 is typical of all the patients in this Group. The initial laterally directed peak reflects the way in which the foot strikes the ground. The medially directed force is of small magnitude in each limb but is usually smaller on the affected side. This is also true for the small laterally directed peak seen immediately prior to toe-off which may represent an attempt to control the transfer of load between the limbs during the phase of double support.

d. Resultant Moments at the Hip and Knee

The moments acting about the joints depend upon the position of the foot relative to the joint and the magnitude of the floor reaction forces.

Hip Joint

The moments about the axis of progression at the hip joint, MHX, can be seen in Figures 4.3a and b. A positive deflection represents a moment which is tending to adduct the hip joint and will therefore be opposed by

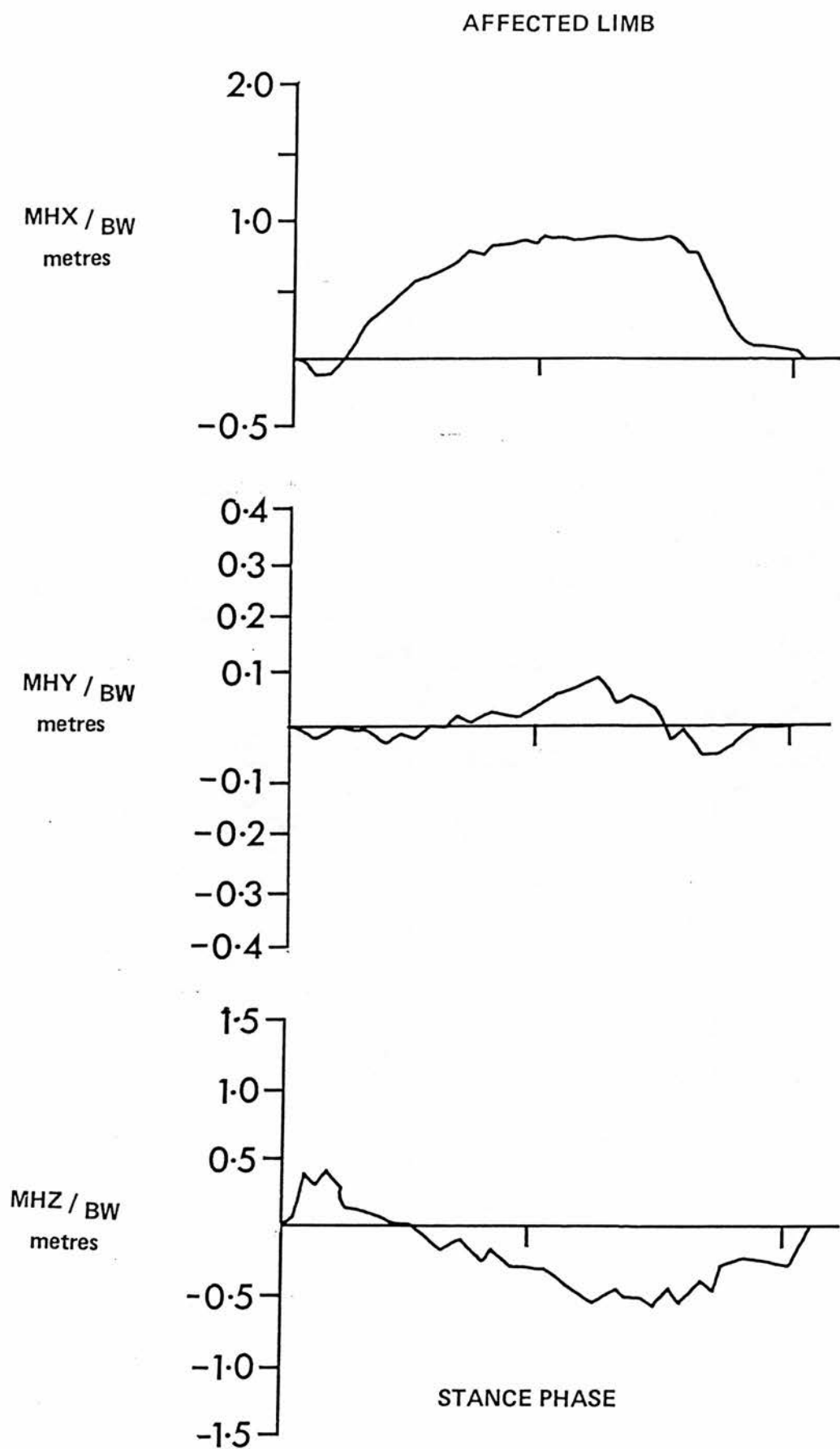


Fig. 4.3a Resultant Moments about the Affected Hip.

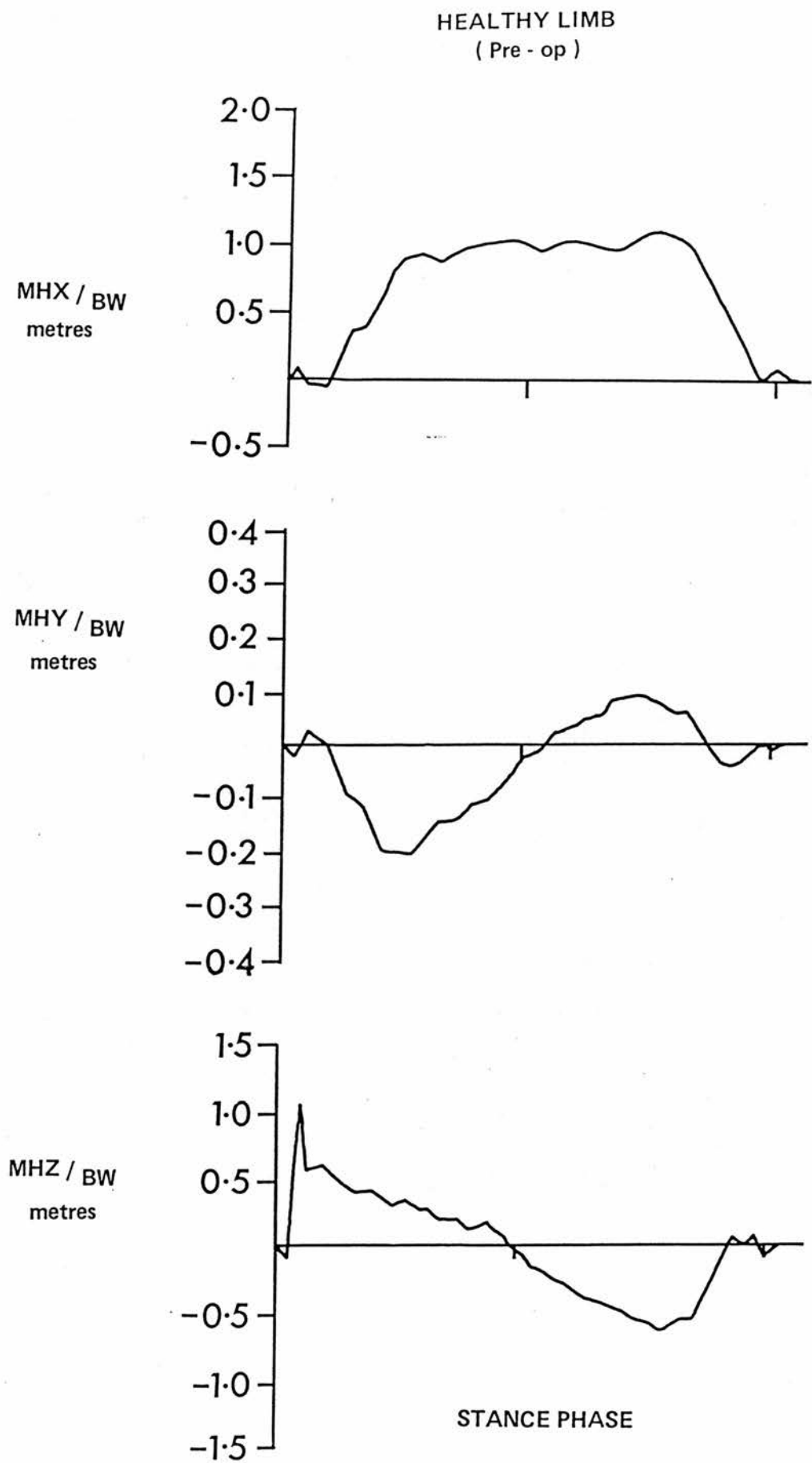


Fig. 4.3b Resultant Moments about the Healthy Hip.

the abductor muscles. The normal pattern has two peaks and a trough but neither limb in this patient, or any of the patients in this Group, exhibit this pattern.

On the affected side there is a slow rise to peak value then a plateau until the moment reduces in magnitude towards toe off. In the healthy limb there is a more rapid rise to peak load and then an irregular plateau. The magnitude of the two curves is approximately the same, although for the Group the peak value of MHX on the affected side is only two thirds of that on the healthy side and both are less than that in the Group of normal subjects (vide infra II.2d).

This patient shows an initial negative deflection in MHX in both limbs (vide infra). This is present in many patients and was noted by McGrouther (1974) who felt it represented a transfer in load from the healthy limb to the affected limb, during the period of double support, by muscle action rather than by the inertial factors which serve this purpose in normal gait.

The moments about the femoral Y axis at the hip joint are small in magnitude on both sides. Their values are very sensitive to marker positioning and therefore liable to a higher degree of error than the other external moments (Chapter 5). However, they do suggest that a lesser degree of rotation takes place about the diseased hip.

MHZ, the flexion-extension moment about the femoral Z axis at the hip joint, reflects the pattern of hip and knee flexion used. On the affected side the initial

deflection indicates a tendency for the hip to flex and must be opposed by the extensor muscles - gluteus maximus and the hamstrings. It is smaller than the deflection for the contralateral limb, and occupies less of the stance phase because less hip flexion is used and with less knee flexion the foot passes behind the hip joint earlier (vide supra). Thereafter the deflection becomes negative indicating a need for flexor muscle activity. In the healthy contralateral limb the pattern is more like that seen in the normal subject and is of similar magnitude.

Knee Joint

The moments about the axis of progression at the knee joint, MKX, (Figs. 4.4a and b) are opposed by force actions in the collateral ligaments. The value and sign of MKX is therefore an indication of the compartmental distribution of the load at the knee joint.

As at the hip joint there is a slower rise to peak value on the affected side than on the healthy side but both curves then show the irregular plateau rather than the peak and trough appearance. This patient therefore transmits most of the load through the medial compartment of each knee. This amount of load transmitted through the medial compartment is variable within the Group (vide infra). The lower value in the healthy limb towards toe-off could be a reflection of the terminal peak in FGZ (Fig. 4.2b) or it may represent the sway of the trunk towards the affected side at this time in the gait cycle. This is not seen in all patients (see Fig. 4.17).

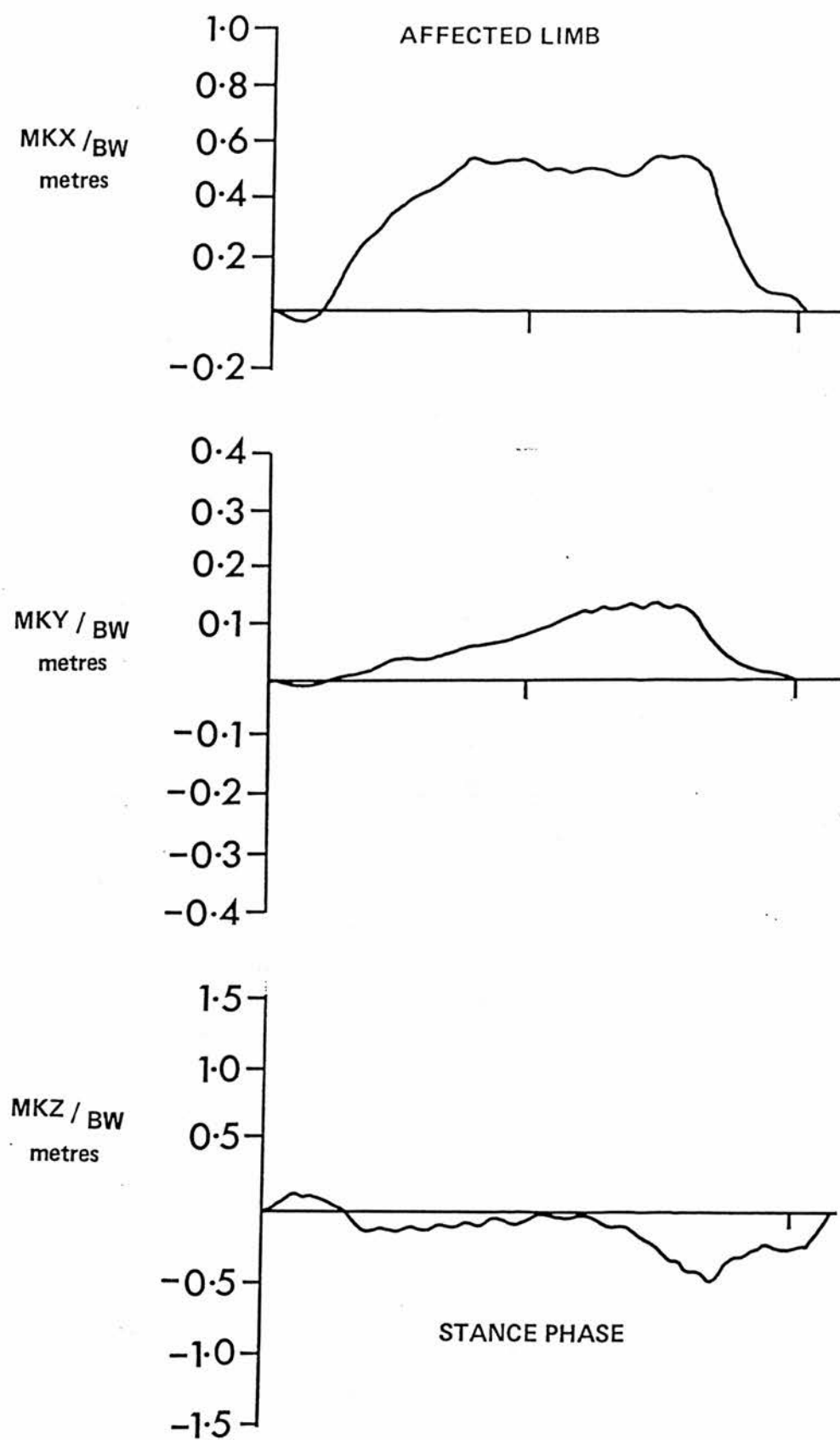


Fig. 4.4a Resultant Moments about the Knee of the Affected Limb.

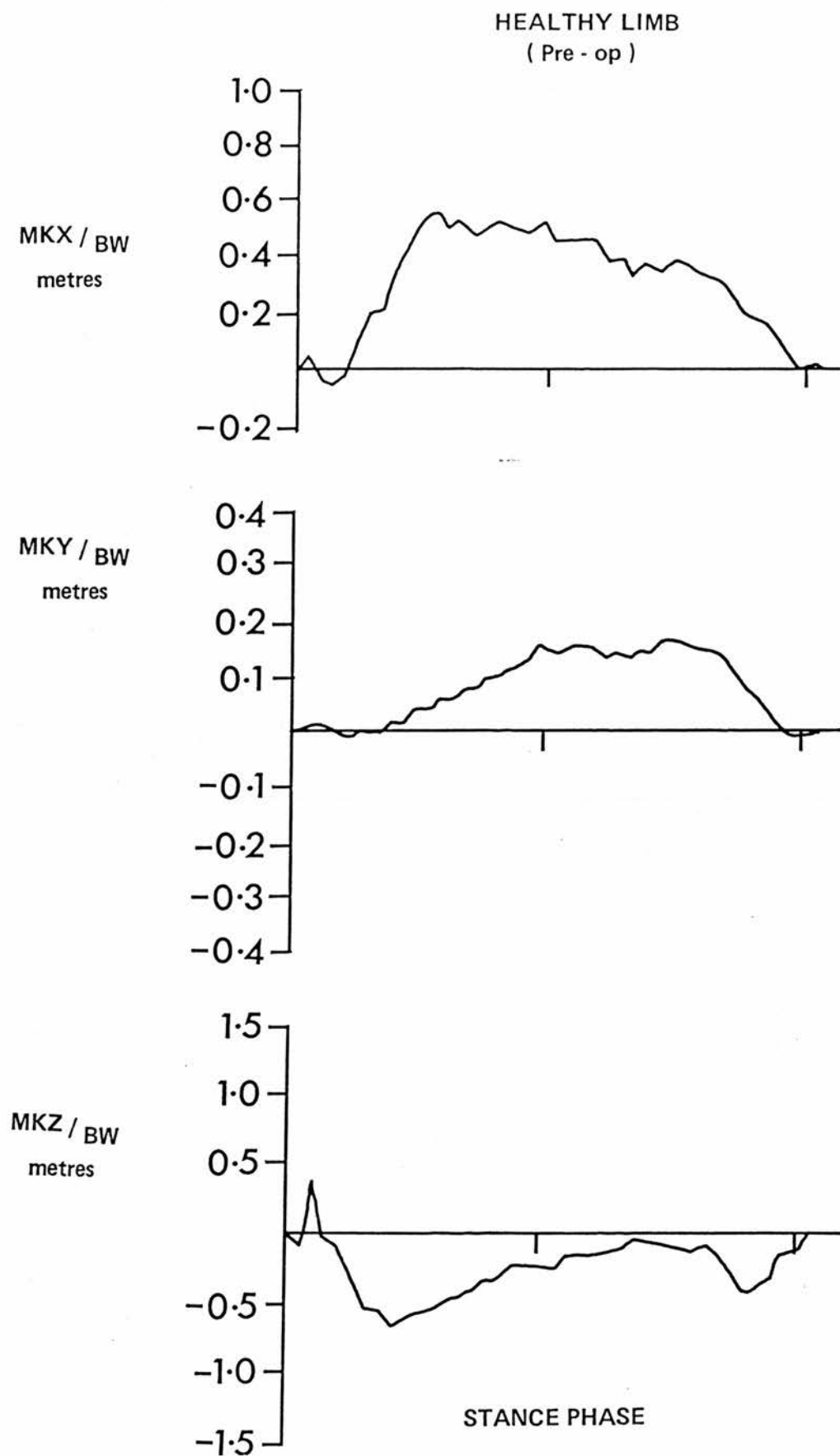


Fig. 4.4b Resultant Moments about the Knee of the Healthy Limb.

The moments about the Y-axis are again small in magnitude but are very similar for each limb.

The flexion-extension moments at the knee, MKZ, reflect the differences in knee flexion seen in each limb; that on the affected side being of small magnitude and becoming negative in late stance indicating a tendency towards flexion which has to be opposed by the knee extensor muscles - quadriceps femoris. On the healthy side the predominantly negative moment reflects the persistent knee flexion.

Muscle and Joint Forces

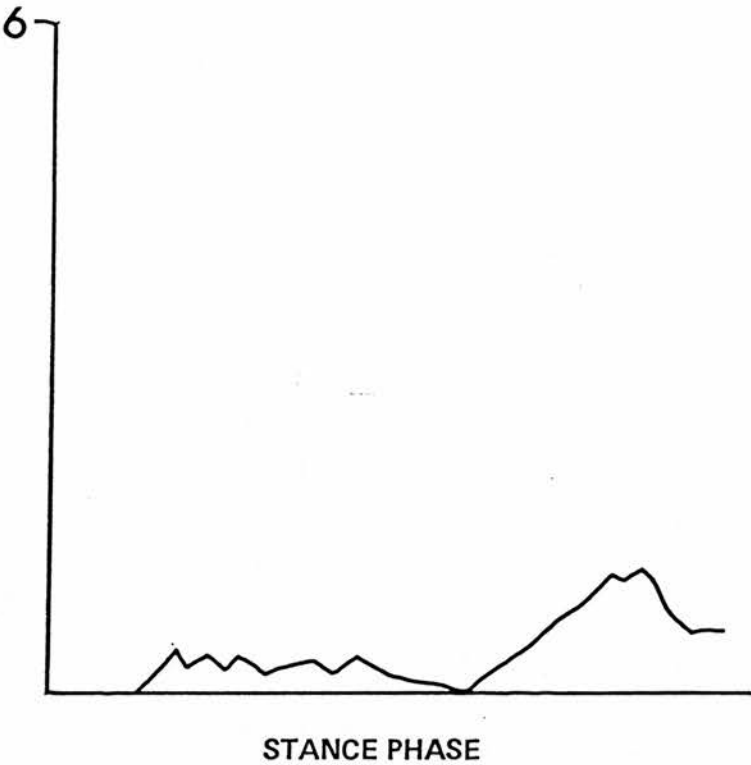
The joint loads are dependent upon the force actions of muscle groups which act to oppose the resultant moments (Chapter 3).

e. Muscle Forces

Knee Joint

At the knee joint the sign of the flexor moment MKZ indicates which muscle groups are active (Chapter 3). In the healthy limb it is predominantly negative indicating that there must be activity in quadriceps femoris to balance the tendency to flexion and activity in this muscle (Fig. 4.5a) is present for all but the beginning of stance phase where MKZ is positive. At this time in the gait cycle the flexor-extensor moment at the hip is also positive indicating that the extensor moment at the knee will be balanced by gastrocnemius or the hamstrings depending upon the relative magnitude of the forces required to balance the Z-plane moments at the

QUADRICEPS
MUSCLE FORCE



QUADRICEPS
MUSCLE FORCE

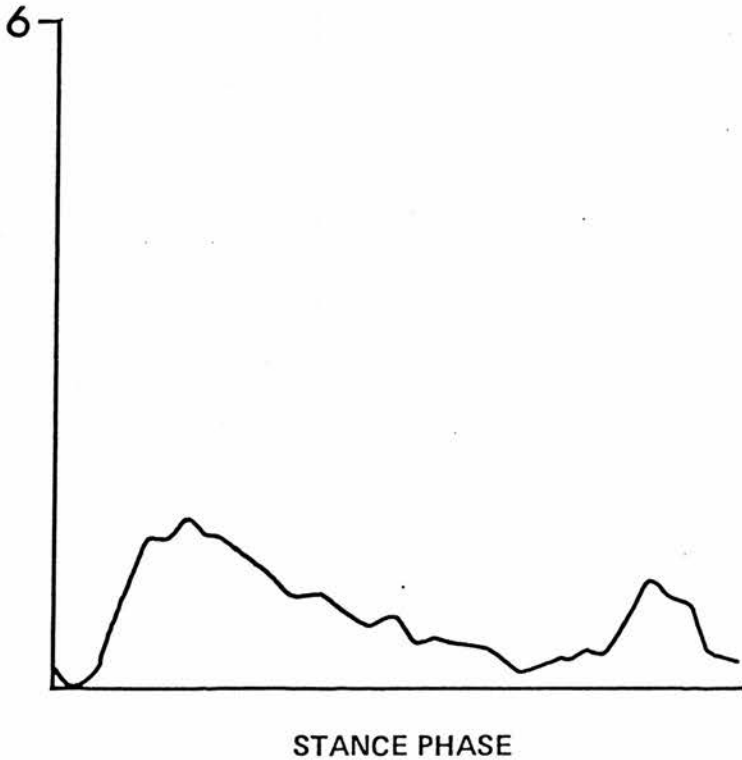


Fig. 4.5a Quadriceps muscle force.

hip and knee. In this case the moment at the hip joint is greater, and the hamstring force would be determined for MKZ leaving the residual moment at the hip to be balanced by gluteus maximus.

In the diseased limb, MKZ is small and variably positive or negative in the first half of stance phase and the low value of quadriceps force results from this. The greater value of quadriceps muscle force in late stance corresponds to the more negative moment in this phase. The very small initial positive moment corresponds to a small hip extensor moment. However, the hip moment is still greater than that at the knee and hamstring force would again be determined by MKZ leaving a smaller residual force at the hip than in the healthy limb to be balanced by gluteus maximus.

Hip Joint

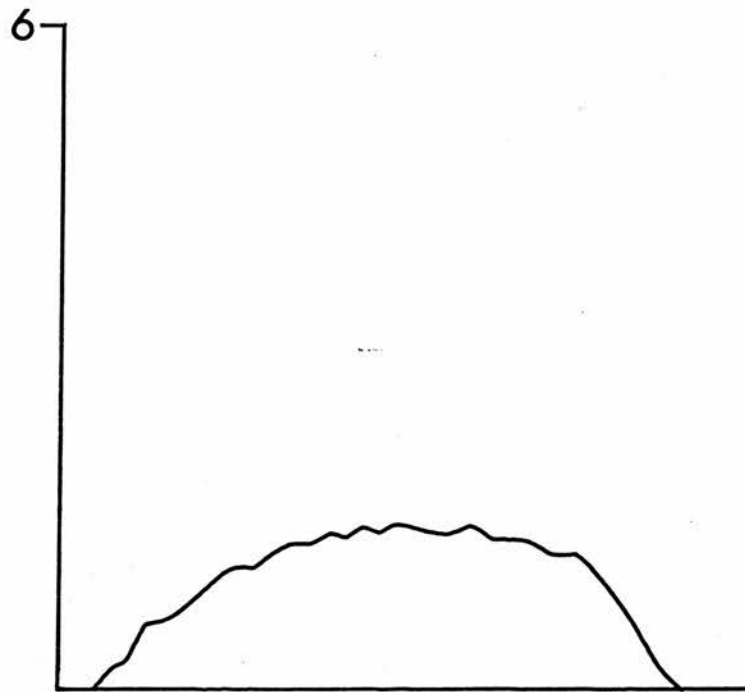
At the hip joint there is an indeterminate problem which has to be solved by the use of a systematic elimination method (Chapters 1 and 3). However, consideration of the moments about the hip joint provide a guide to which muscles are active.

MHX acts throughout stance phase and is positive in both limbs implying force in the abductor muscles and this is seen in Figure 4.5b, differences in the force magnitudes mirroring the differences in magnitude of MHX.

MHZ differs in magnitude and form between the two limbs. In the first part of stance phase on the healthy side the moment is positive and large compared to that

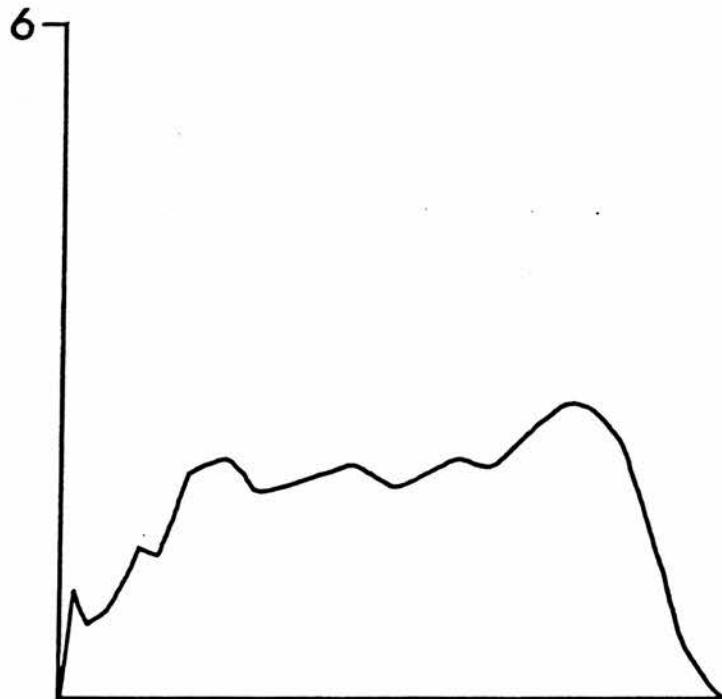
AFFECTED LIMB

ABDUCTOR
MUSCLE FORCE



HEALTHY LIMB
(Pre - op)

ABDUCTOR
MUSCLE FORCE



STANCE PHASE

Fig. 4.5b Hip abductor muscle force.

on the diseased side. It also occupies a greater proportion of stance phase. This is reflected in the greater magnitude and duration of extensor muscle force seen on the healthy side. As already mentioned, this will be gluteus maximus on both sides. The value for the total extensor muscle force at the hip joint is shown in Figure 4.6..

In the later part of stance phase MHZ is negative in both limbs requiring activity in the hip flexors to balance it (Fig. 4.6). The difference in magnitude and duration of the flexor muscle force corresponds to the pattern of the moments.

f. Joint Loads

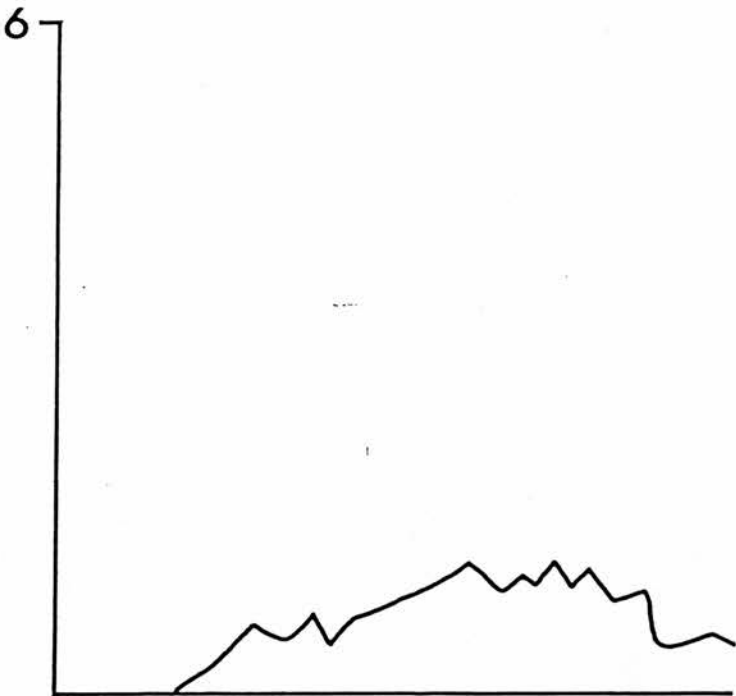
Knee Joint

The total knee joint load is shown in Figure 4.7, and it can be seen how closely its shape resembles that of the quadriceps femoris force, reflecting the importance of this muscle group in the generation of load across the knee. However, the pattern differs from that seen in the normal knee joint (Morrison, 1967) in both limbs. Morrison identified three peaks in the load/time curve for the knee joint (Fig. 1.10), 'a' corresponding to hamstring action, 'b' corresponding to quadriceps femoris and 'c' corresponding to gastrocnemius, separated from 'b' by a trough.

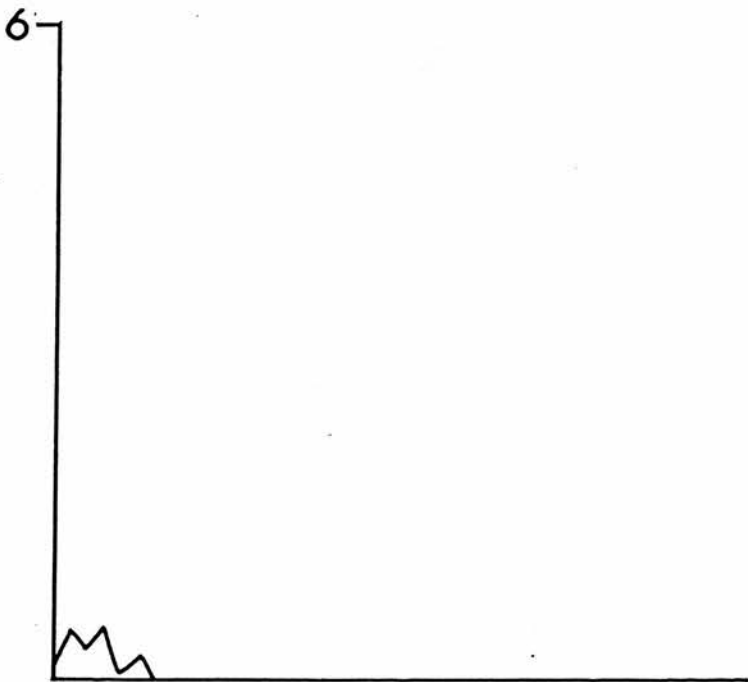
In the healthy limb peak 'a' is similar to that for the group of normal subjects, peak 'b' is of similar magnitude to normal (2.7xBW) but peak 'c' is considerably

AFFECTED LIMB

HIP FLEXOR
MUSCLE FORCE



TOTAL
EXTENSOR MUSCLE
FORCE



STANCE PHASE

Fig. 4.6a Hip Flexor and Extensor Muscle Force - Affected Limb.

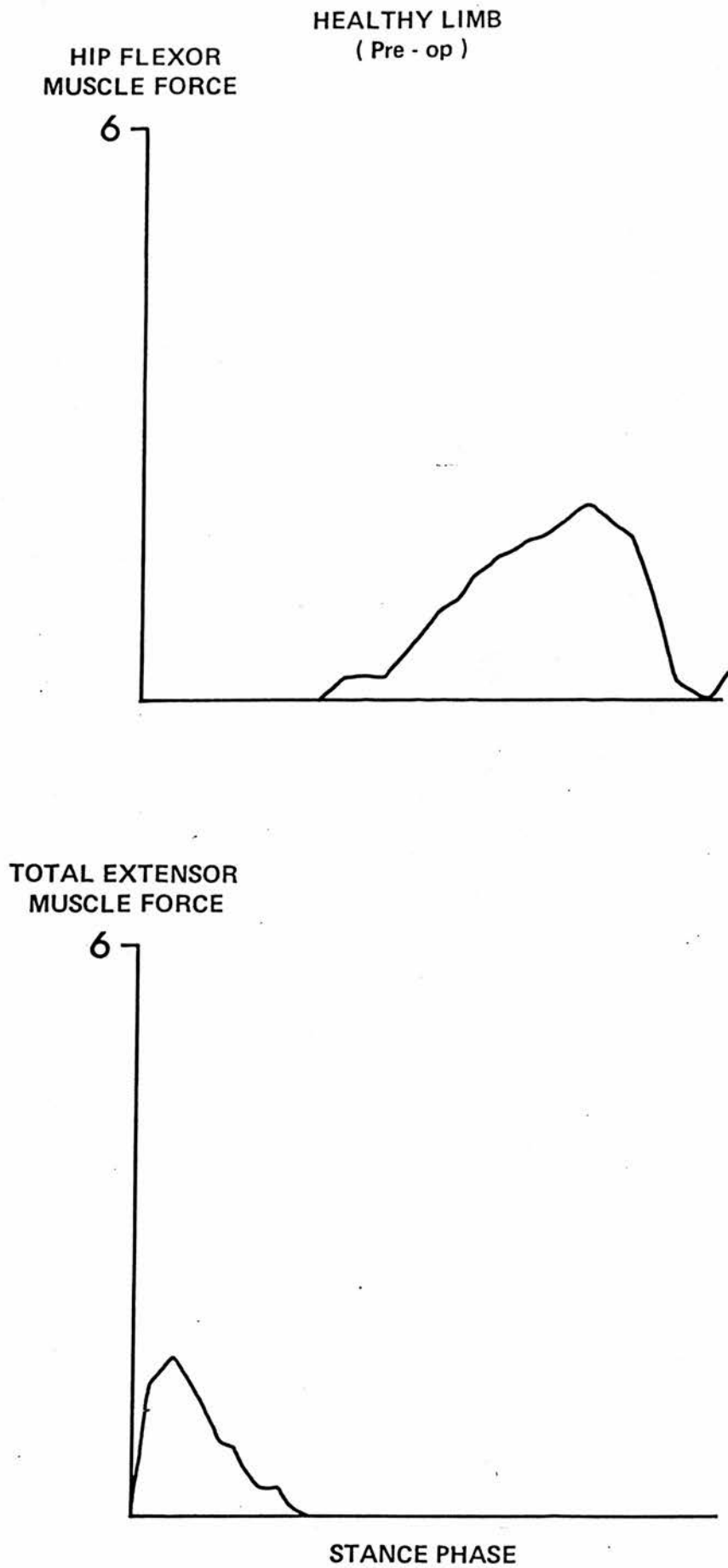


Fig. 4.6b Hip Flexor and Extensor Muscle Force
- Healthy Limb.

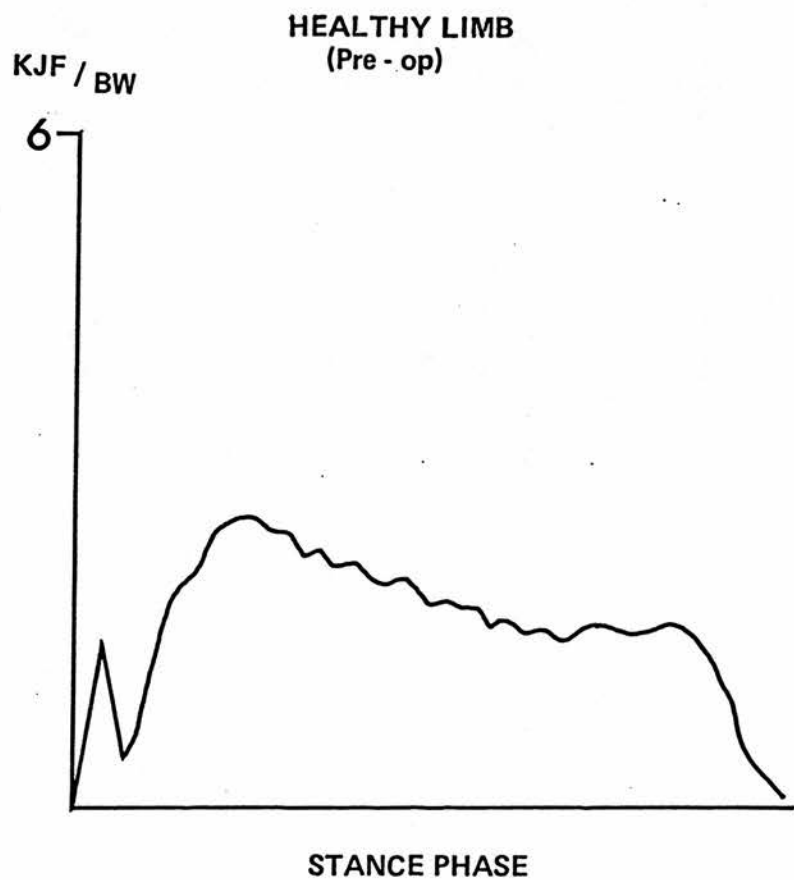
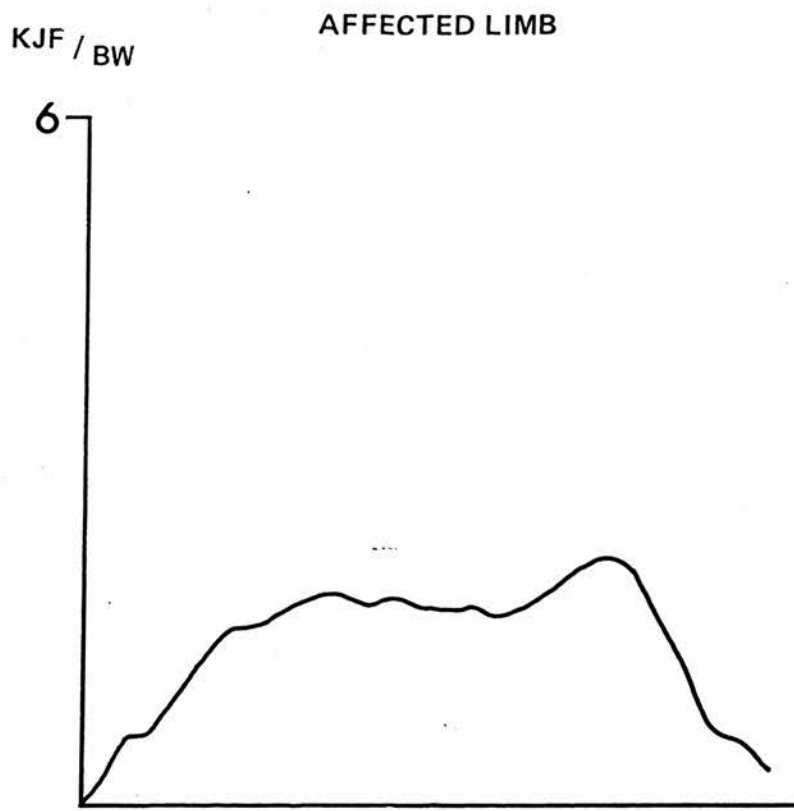


Fig. 4.7 Knee Joint Load.

less than normal ($1.8 \times BW$) and no trough is seen. This corresponds to the small value of MKZ at this phase of stance and indicates that there was less gastrocnemius activity.

On the ipsilateral diseased side again the three peaks can be discerned but here the 'b' peak is small because of the low force in the quadriceps acting about the nearly fully extended knee.

The compartmental distribution of the load is seen in Figures 4.8a and 4.8b. From the sign of MKX it was clear that the major part of the load in both limbs was transmitted across the medial knee joint compartment and this is confirmed. Indeed, very little load is transmitted in the lateral compartment in either knee. However, some patients showed a greater proportion of load being carried by the lateral compartment of the knee on the affected side (vide infra).

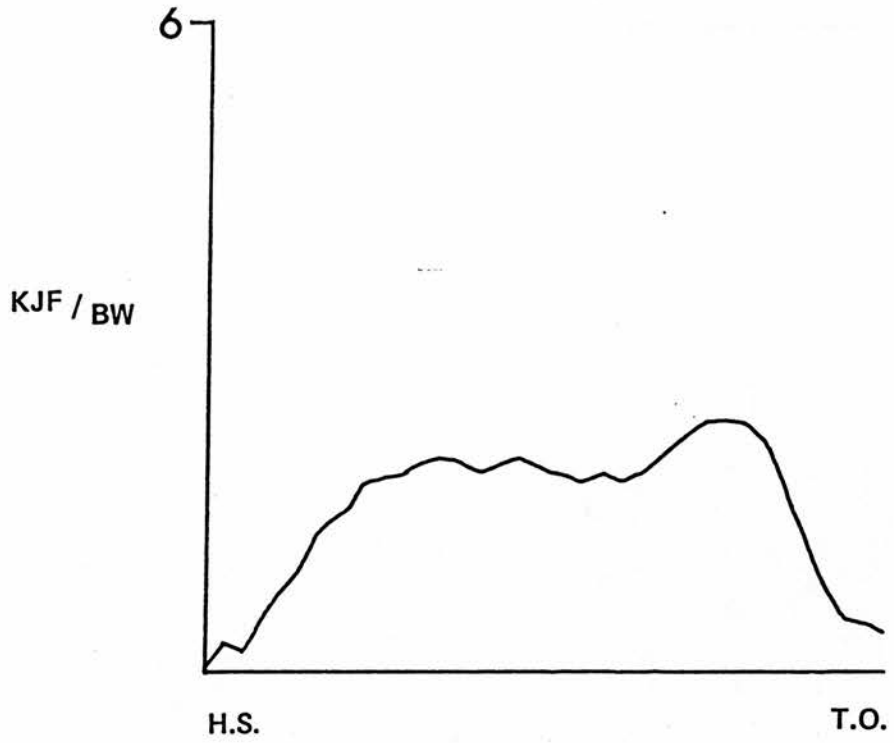
Hip Joint

The resultant hip joint force is shown in Figure 4.9. Differences in pattern and magnitude are seen between the two hips and for the same hip on the healthy side.

On the healthy side there is a small heel strike peak and a larger toe-off peak (3.3 and $5.2 \times BW$). The first peak arises from force in the abductors and hip extensor muscles and the second peak results from hip flexor activity made necessary by the extension range used.

AFFECTED LIMB

MEDIAL
JOINT COMPARTMENT



LATERAL
JOINT COMPARTMENT

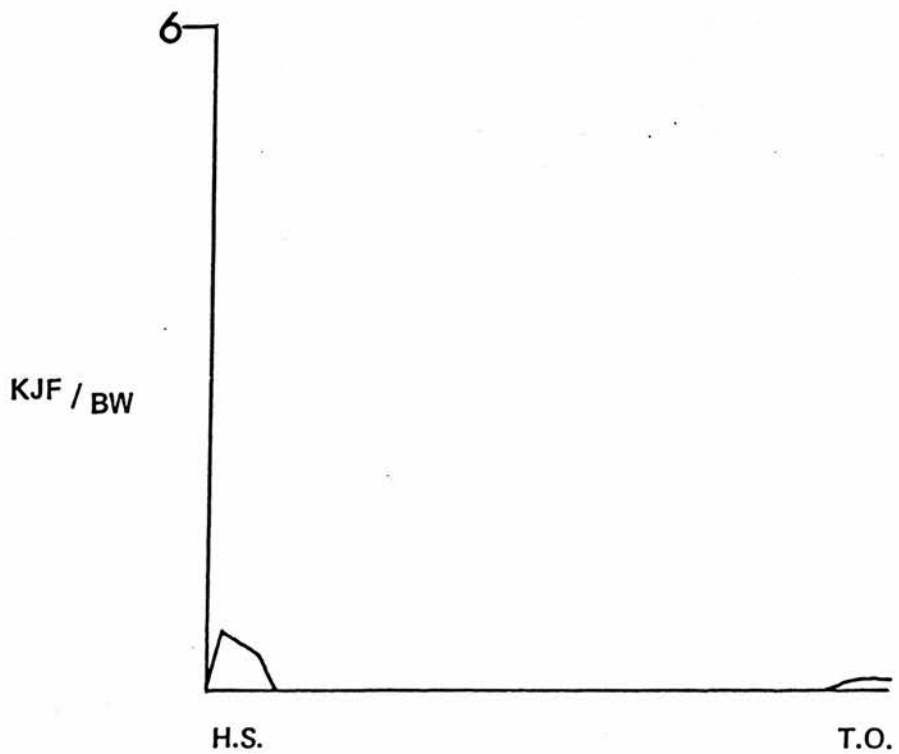


Fig. 4.8a Compartmental knee joint loads - affected limb.

HEALTHY LIMB
(Pre - op)

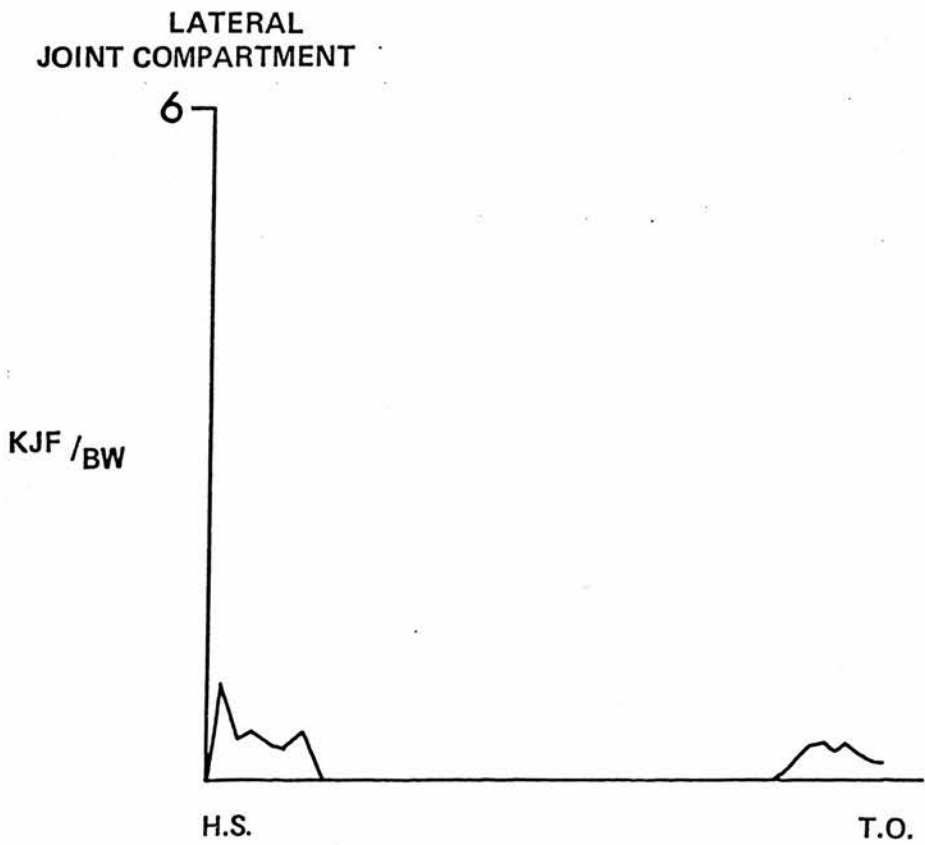
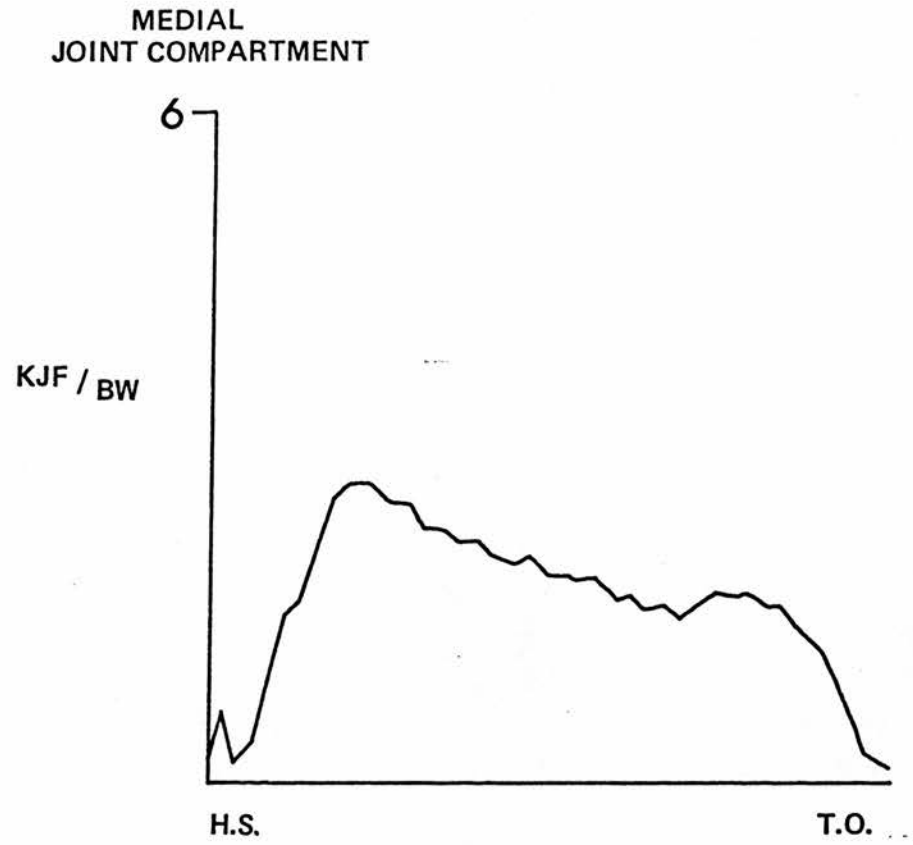
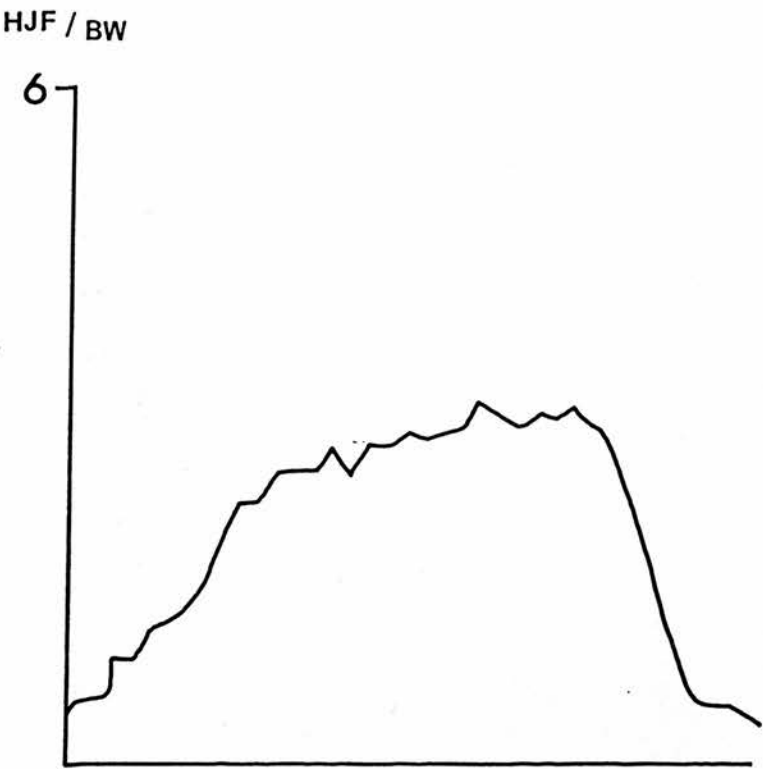
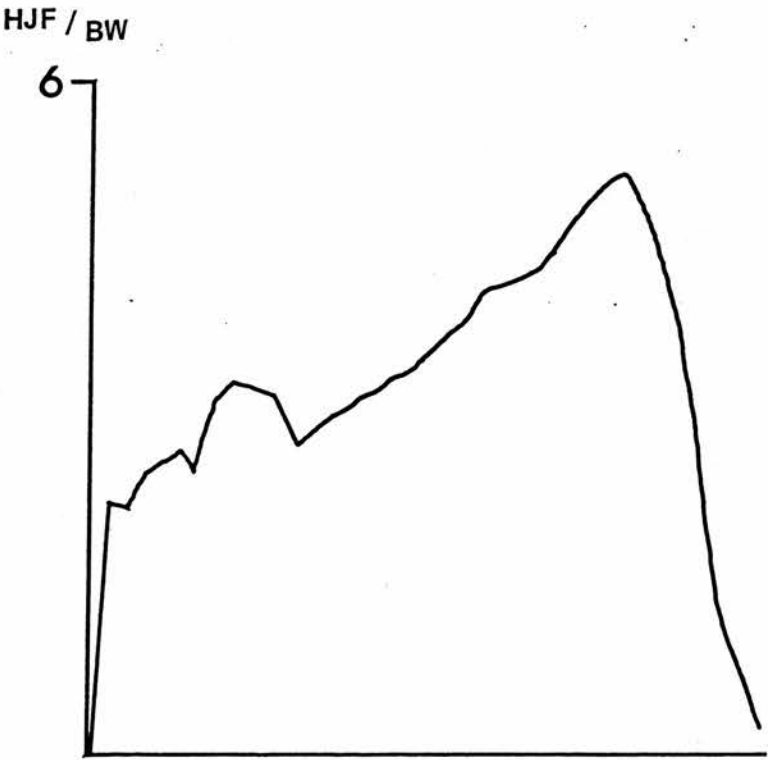


Fig. 4.8b Compartmental knee joint loads -
healthy limb.

AFFECTED LIMB



HEALTHY LIMB
(Pre - op)



STANCE PHASE

Fig. 4.9 Hip Joint Load

The diseased hip shows an almost pointed curve of lesser magnitude than either of those on the healthy side ($2.7 \times BW$ in the first part of stance and $3.0 \times BW$ towards toe-off). This pattern mirrors that of the moments and the low force in the first part of stance is due to the low abductor and extensor forces. Activity in the flexors commences early in stance because of the limb movement pattern, the limb passing into extension before 50% stance and together with the abductors is responsible for the peak of the curve.

4.II

2. Summary of Results for Hip Disease Group

a. Time-Distance Parameters

All patients walked at a slower speed than the normal subjects ($0.83 \pm 0.54 \text{ m.s}^{-1}$) and all had a reduced duration of stance phase on the affected limb when compared to the healthy limb (0.74 ± 0.24 compared to 0.82 ± 0.21 sec; $p = 0.001$ -paired t-test).

b. Limb Movements

There was a large scatter in the range of hip and knee movements (Figs. 4.10 and 11) but consistent differences in the total range and pattern of movements were seen between the limbs. There was less flexion and usually less extension at the affected hips. At the knees of the affected limbs there was a loss of the normal triphasic pattern in nearly all whereas this was preserved in the majority of the contralateral knees. The pattern at the affected knee was typically that of remaining in a small degree of flexion until double support phase began and flexion commenced. The majority of the contralateral knees, whilst retaining the normal pattern, exhibited more flexion (in the region of 30° compared to the normal group's mean of 22°) in early stance, and failed to extend as far as the normals at mid-stance.

c. Ground-Foot Reaction Forces

A much smaller scatter of values is seen when the ground-foot forces are considered for the group (Figs. 4.12 and 4.13), with the exception of the results for FZG.

AFFECTED LIMB

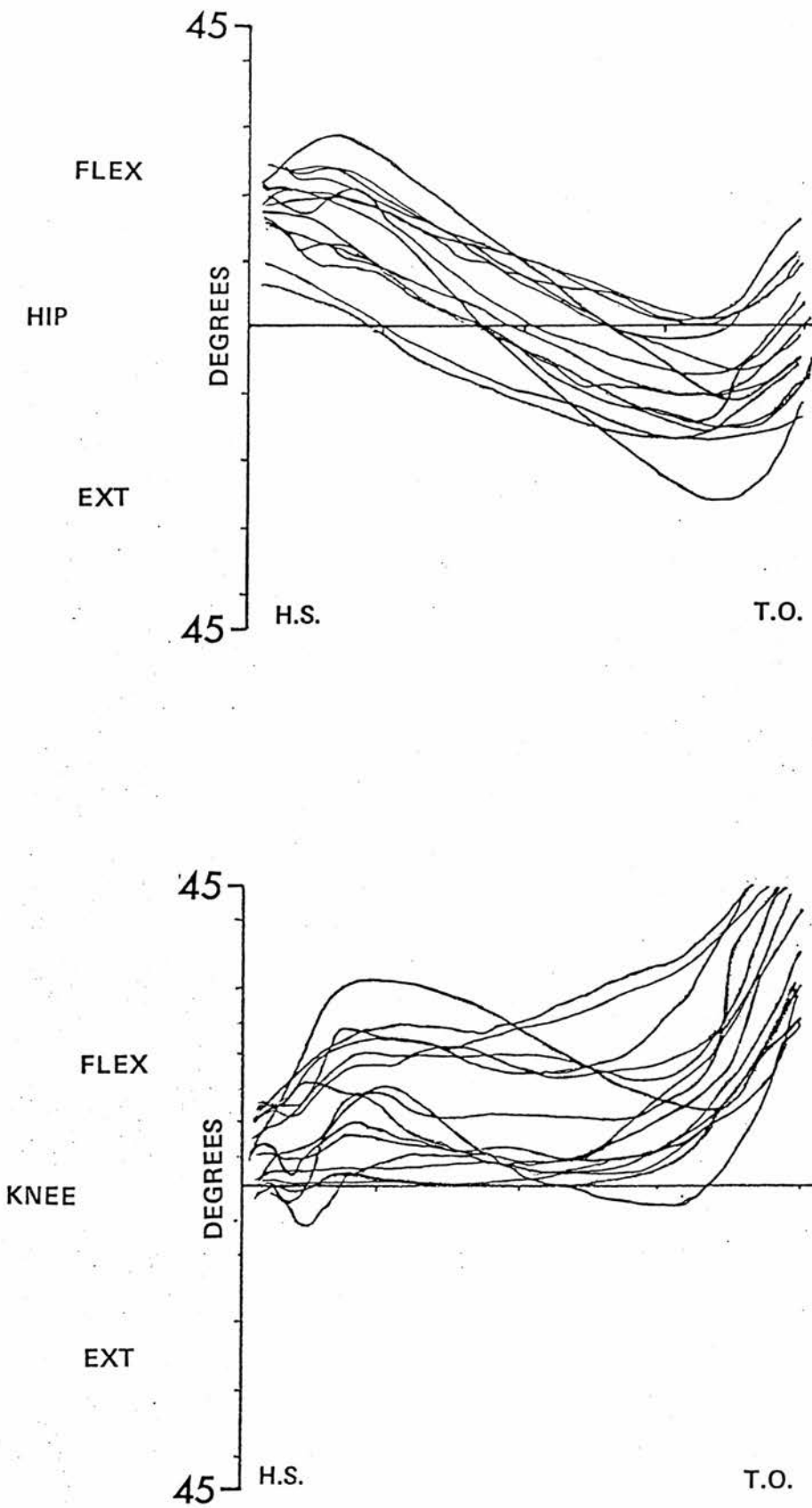


Fig. 4.10 Range of movement at hip and knee joints
- affected limbs.

HEALTHY LIMB
(Pre - op)

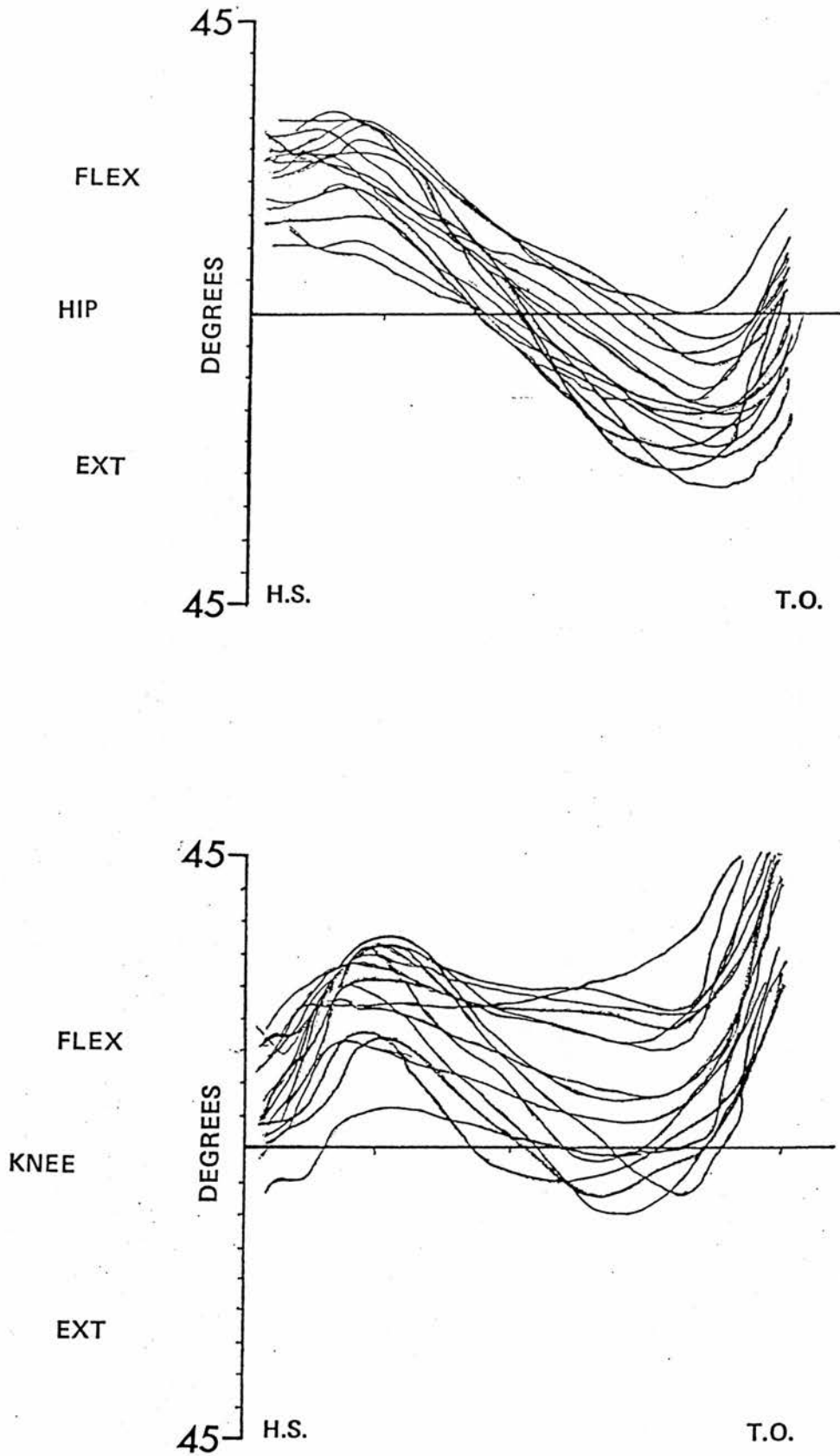


Fig. 4.11 Range of movement at hip and knee joints
- healthy limbs.

AFFECTED LIMB

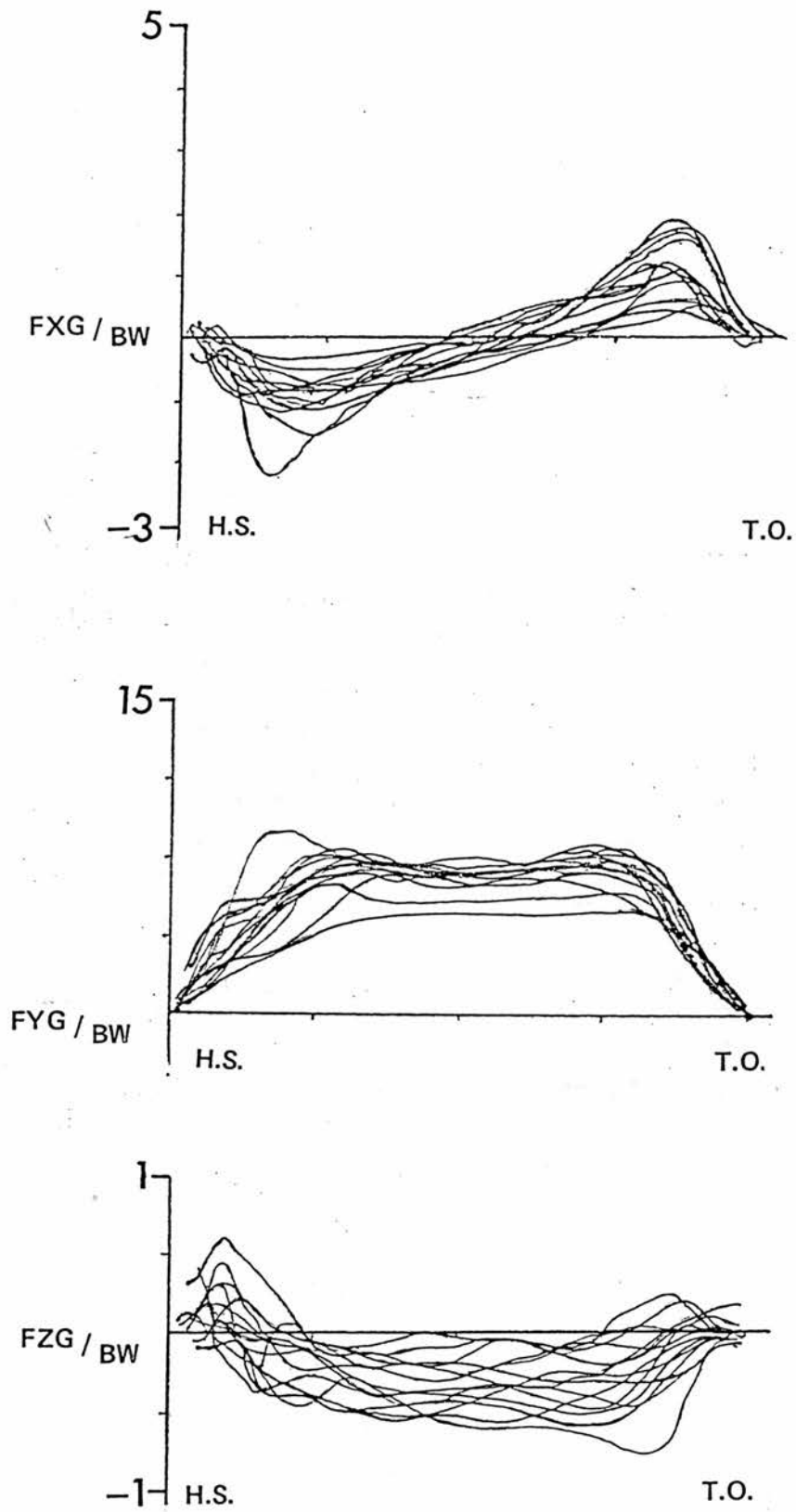


Fig. 4.12 Ground Reaction Forces
- Affected Limbs.

HEALTHY LIMB
(Pre - op)

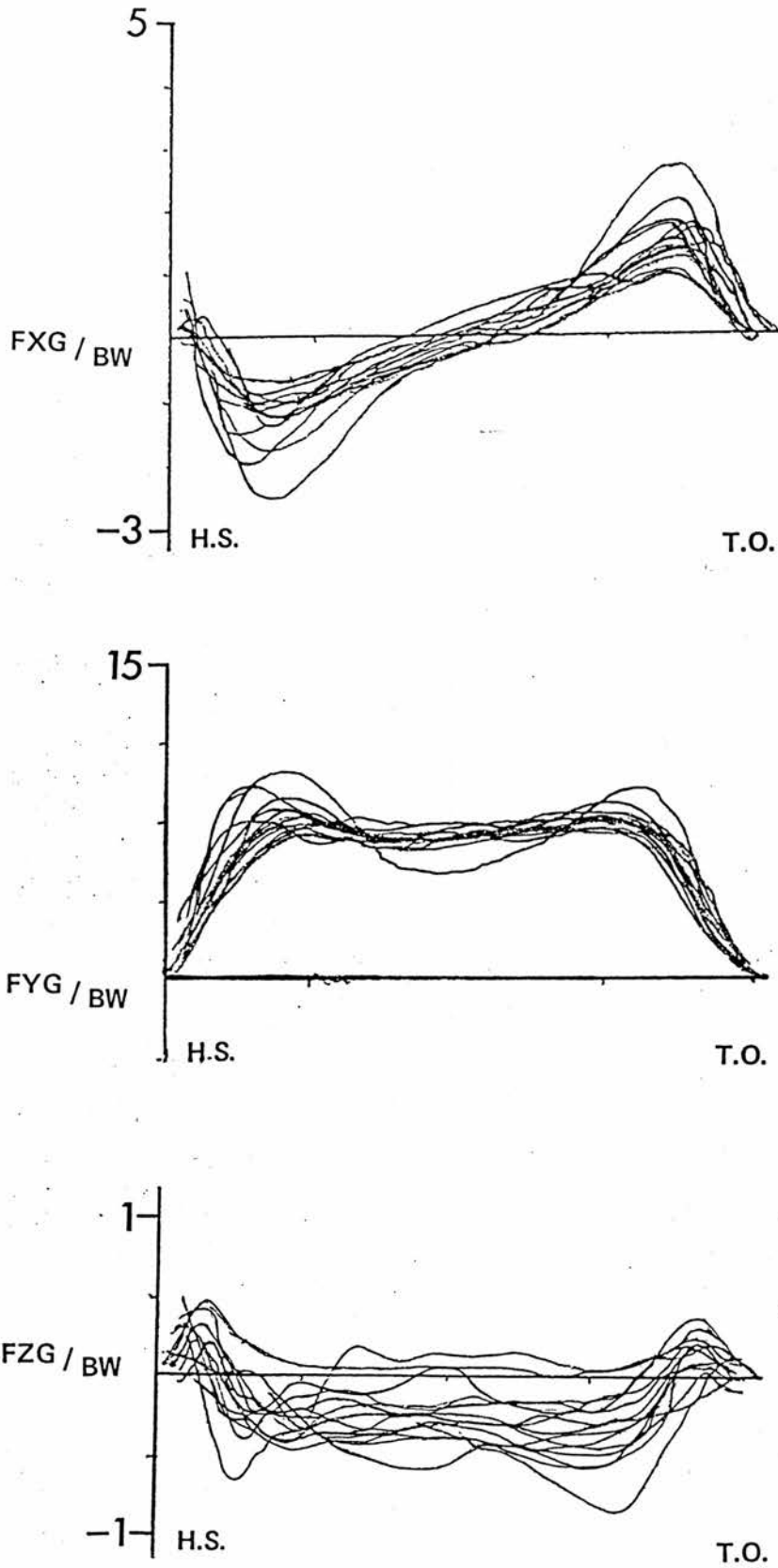


Fig. 4.13 Ground Reaction Forces - healthy limbs.

The pattern of FXG is the same for both limbs there being a greater magnitude of retarding and propulsive forces on the healthy limbs.

FYG is similar in magnitude for both limbs, and in both limbs it lacks the typical two peak and trough pattern. There is a less rapid rise to the first peak in the affected limbs, suggesting a slower rate of vertical loading.

Despite the variability in FZG the pattern is similar for both limbs with an initially laterally directed force followed by a medially directed force for most of stance phase and with a terminal laterally directed force immediately before toe-off. This last peak is frequently absent in the affected limbs. Its possible reflection of the active transfer of load between the limbs in the phase of double support has already been mentioned (vide supra).

d. Resultant Moments at the Hip and Knee

Hip Joint

The very great difference between the limbs in the pattern of the curve for MHX, the moment about the axis of progression, is apparent from Figures 4.14 and 4.15. On the affected side there is universal loss of the two peak and trough pattern and in two cases the value of the moment is close to zero. The typical pattern is replaced either by an irregular plateau or by a rise to a single peak, close to the time of mid-stance.

In the healthy limbs the normal pattern is retained by a few patients but the majority show an irregular plateau.

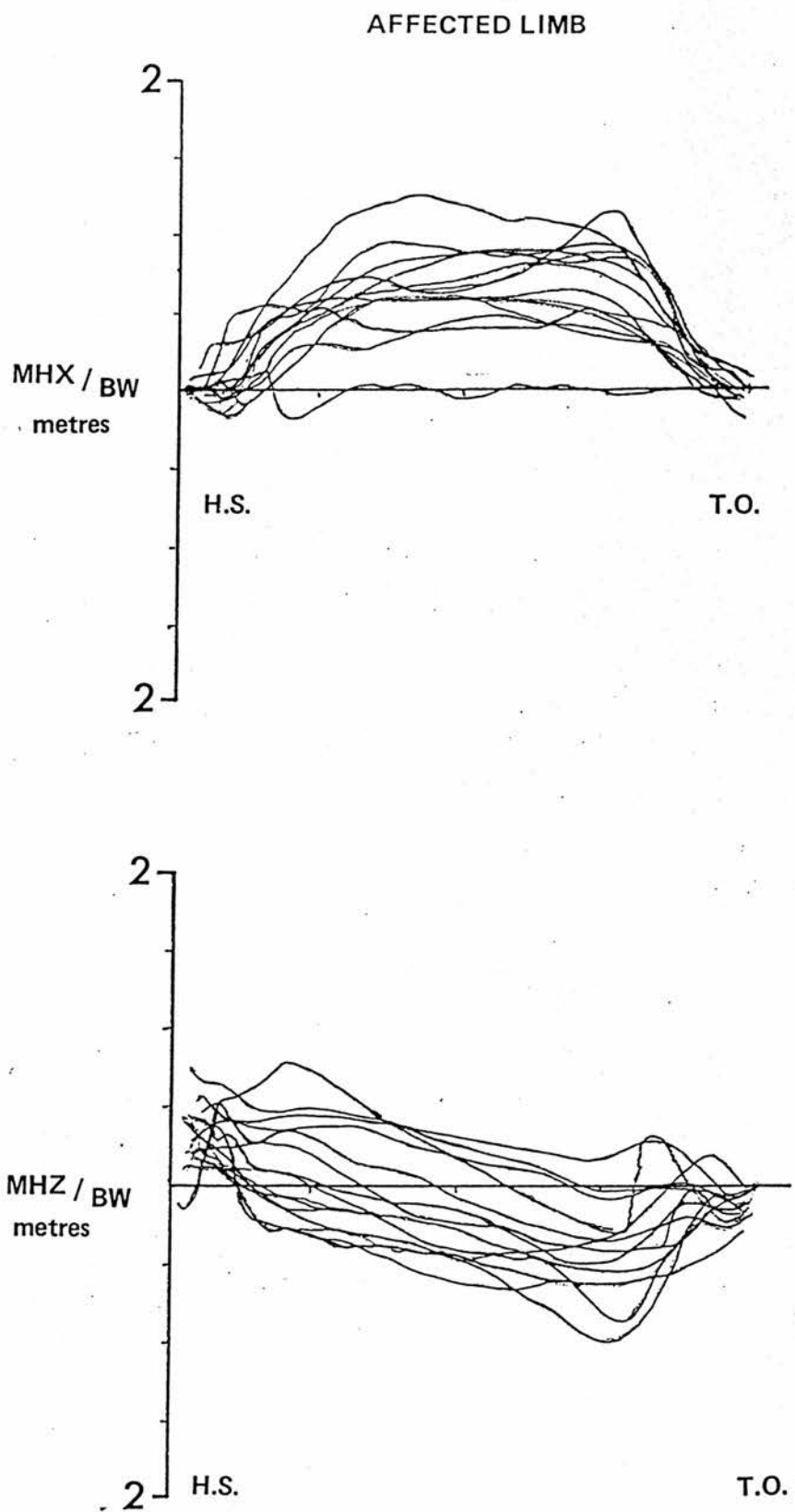


Fig. 4.14 Resultant moments about the hip joints
- affected limbs.

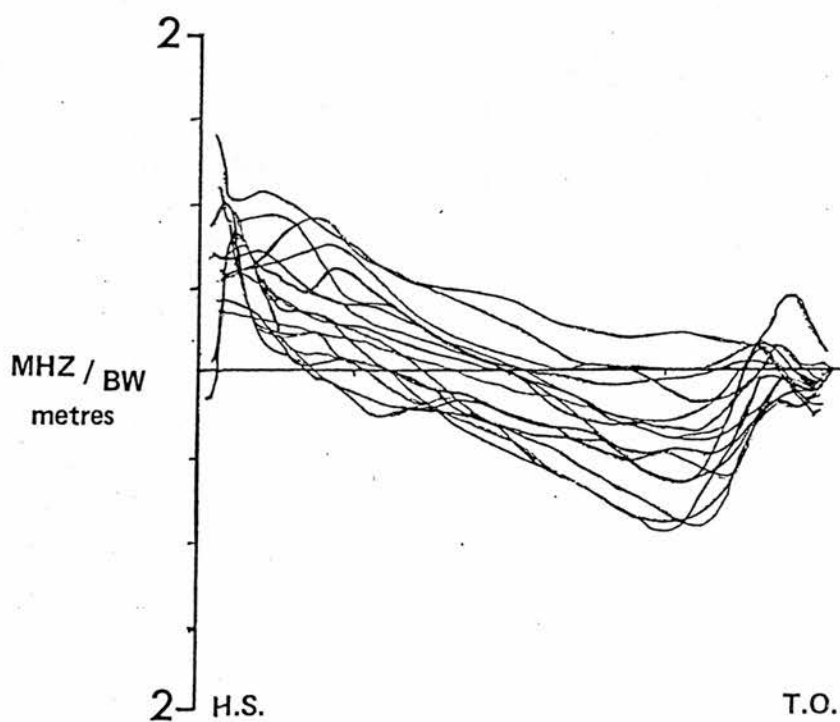
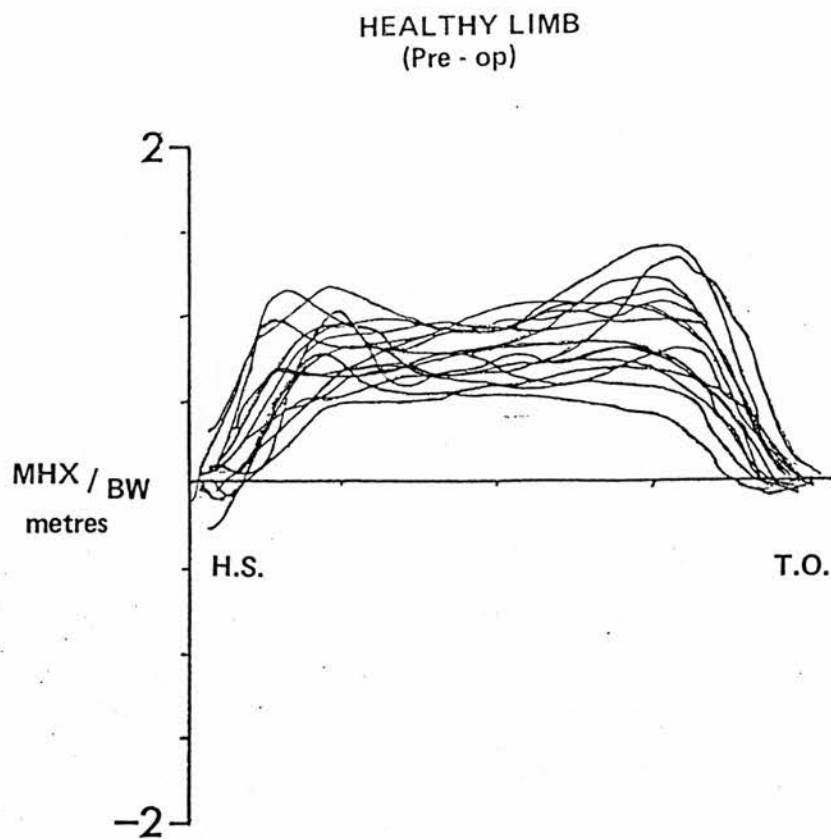


Fig. 4.15 Resultant moments about the hip joints - healthy limbs.

The mean values for the peak moment for the two limbs show a difference of some 30 per cent between the limbs (healthy limb 0.88 ± 0.42 metres; affected limb 0.62 ± 0.54 metres) but this is heavily influenced by the two patients with extremely small moments acting about their affected hip joints.

The moments about the vertical axis (MHY) are not considered in the later processing of results and will not be considered here.

MHZ, the flexion-extension moment, exhibits extreme variability in both magnitude and pattern with only very small differences between the limbs.

Knee Joint

As previously mentioned MKX gives an indication of the compartmental load distribution at the knee joint. In this group (Figs. 4.16 and 4.17) both limbs usually show an irregular plateau rather than the two peak and trough configuration of the normal subject. However, in the healthy limbs this can occasionally be discerned. The magnitude of the moment is greater in the healthy knees indicating a greater proportion of the load is being transmitted through the medial joint compartment in these limbs. The sign and value of the moment on some of the affected limbs suggests a tendency to a greater transmission of load through the lateral compartment. This might result from severe trunk sway.

The marked initial negative deflection seen in most of the healthy limbs again probably represents both the action of the foot in striking the ground (See Fig. 4.13 FZG)

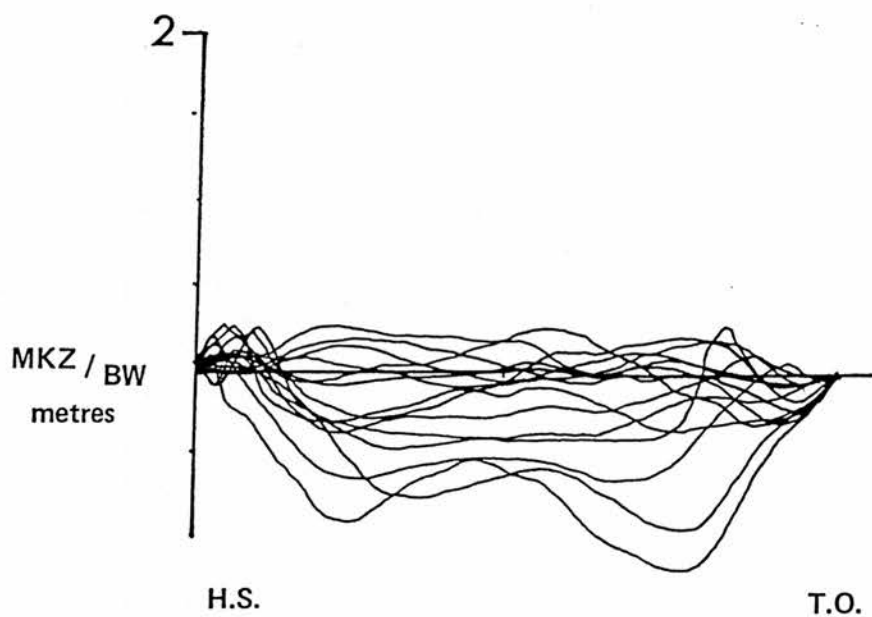
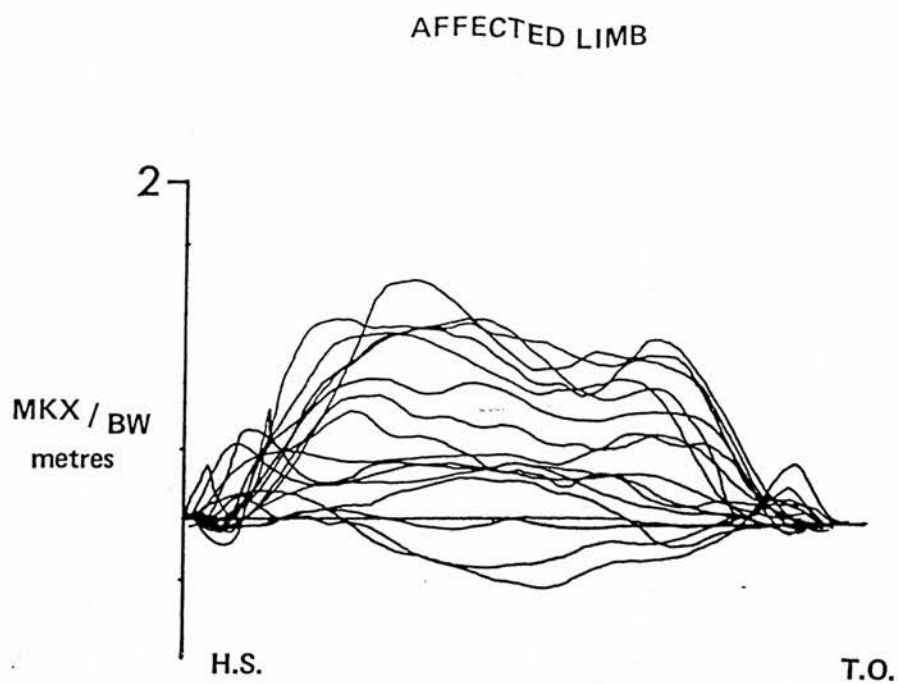


Fig. 4.16 Resultant moments about the ipsilateral knee joints.

HEALTHY LIMB
(Pre - op)

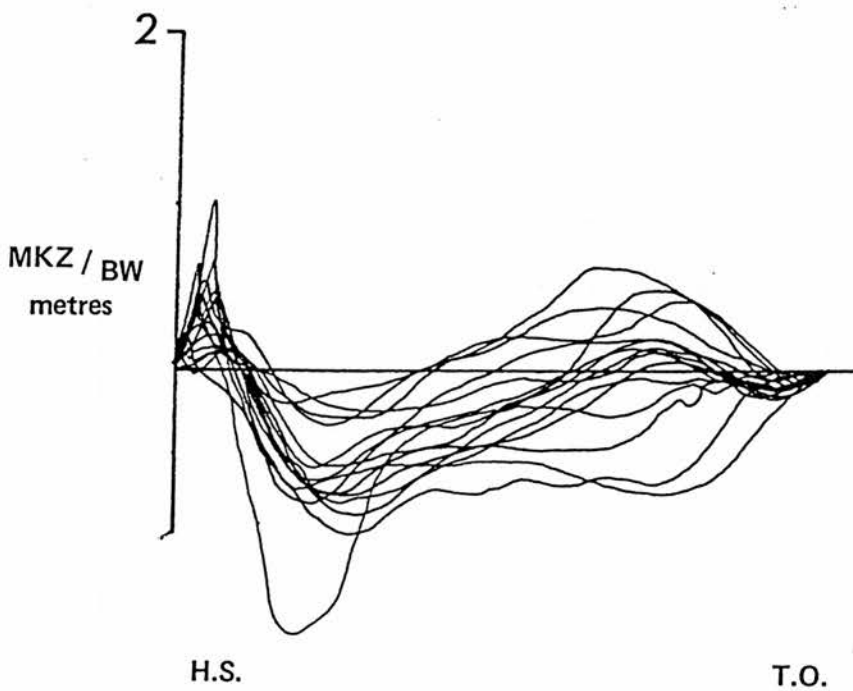
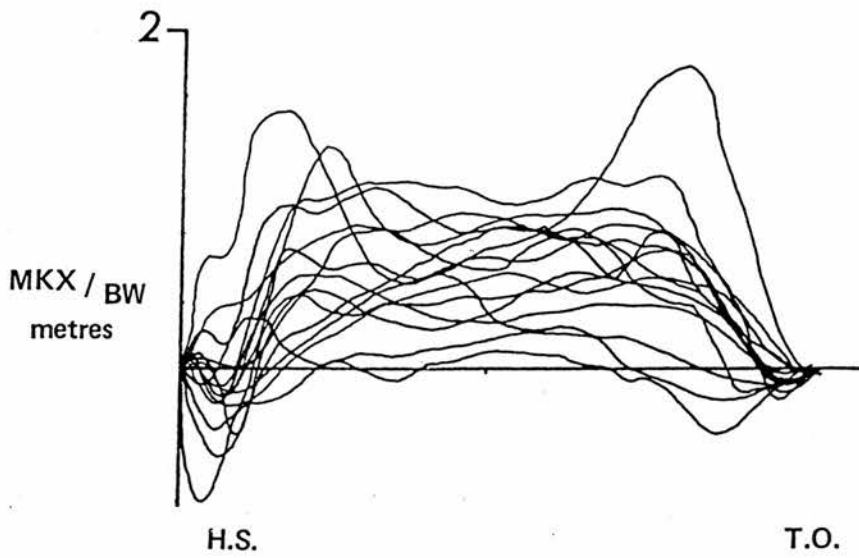


Fig. 4.17 Resultant moments about the
contralateral knee joints.

and the position of the trunk relative to the foot at heel strike.

MKZ provides an indication of action in quadriceps femoris, gastrocnemius and the hamstrings. Differences between the limbs are apparent from Figures 4.16 and 4.17. The predominate sign in both limbs is negative indicating the need for action in quadriceps femoris. In most of the affected knees this is the situation throughout stance phase reflecting the stiffly held knee in that limb (see Fig. 4.10). Some affected side knees had virtually no flexor moment. In some of the contralateral healthy knees quadriceps femoris action was necessary throughout stance to balance the moment but in several the moment became positive in sign during the second part of stance phase reflecting the retention of the triphasic pattern of movement and indicating the need for gastrocnemius action (see Fig. 4.11 and vide infra).

The initial positive peak seen in the traces for both limbs, but more marked in the healthy limbs, indicates the force of the heel strike and indicates the need for hamstrings action at the knee (vide infra).

e. Muscle Forces

The peak values for forces in the hip abductor, flexor and extensor muscles and in quadriceps are shown for each patient in Table 4.2. At the hip joint some patients show little activity in the abductors and much in the hip extensors or flexors whilst others show pre-dominant use of the abductors or a more "normal" combination of both. The same pattern of muscle usage is usually seen in each limb.

The magnitude of the muscle forces at the affected hip is much less than those at the healthy hip and this is particularly notable when the abductor muscles are considered (1.38 ± 1.42 xBW for the affected hip and 2.52 ± 1.68 xBW for the healthy hip; $p = 0.001$ (paired t-test)).

f. Joint Loads

Knee Joint

The curves for the loads across both knee joints of the pre-operative group are seen in Figures 4.18 and 4.19. In the contralateral knee the three peak pattern is usually seen although the terminal peak, corresponding to gastrocnemius contraction is frequently small. In the ipsilateral knee the most usual form of curve is an irregular plateau although more typical curves are seen in a few patients.

The mean knee joint loads for the group are 2.23 ± 1.20 xBW for the ipsilateral knees and 2.77 ± 1.78 xBW for the contralateral knees ($p = 0.1$ paired t-test).

The compartmental loading patterns are very valuable (Table 4.2). In the ipsilateral knees 26.06 ± 8.98 per cent of the total knee joint load was transmitted by the lateral joint compartment compared to 23.43 ± 6.68 per cent in the contralateral knees. It can be seen from Table 4.2 that both extremes of compartmental loading were seen.

Hip Joint

The hip joint loads for the group are shown in Figures 4.20 and 4.21. In the healthy limbs the forces are usually of greater magnitude (4.83 ± 2.04 xBW for the

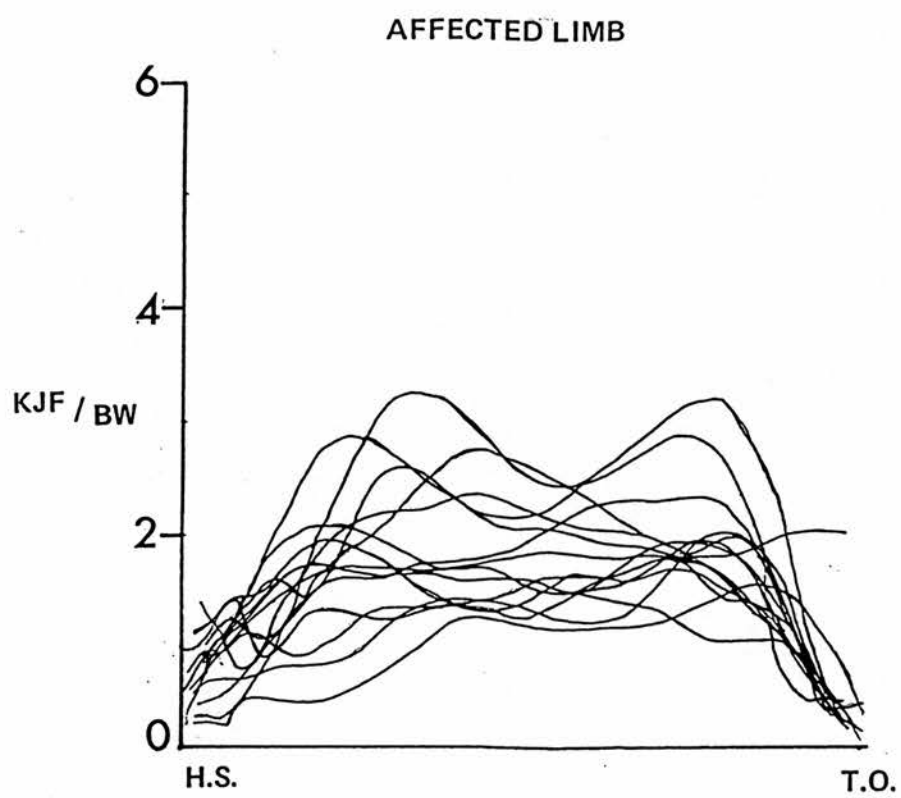


Fig. 4.18 Knee Joint Loads - Ipsilateral Knees.

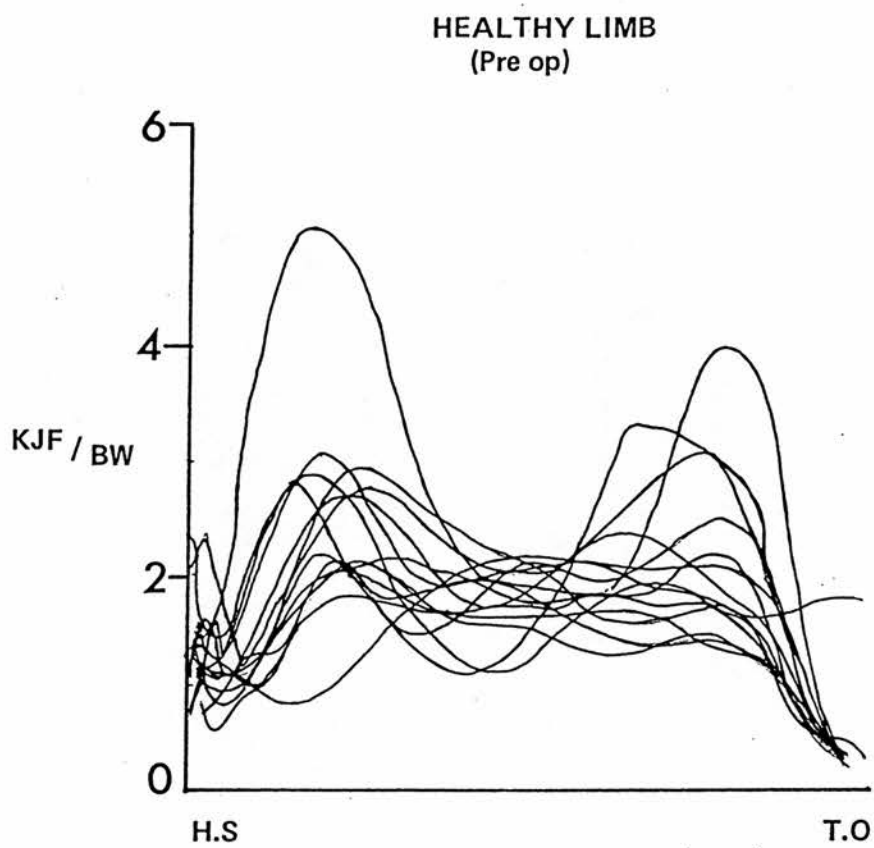


Fig. 4.19 Knee Joint Loads - Contralateral Knees.

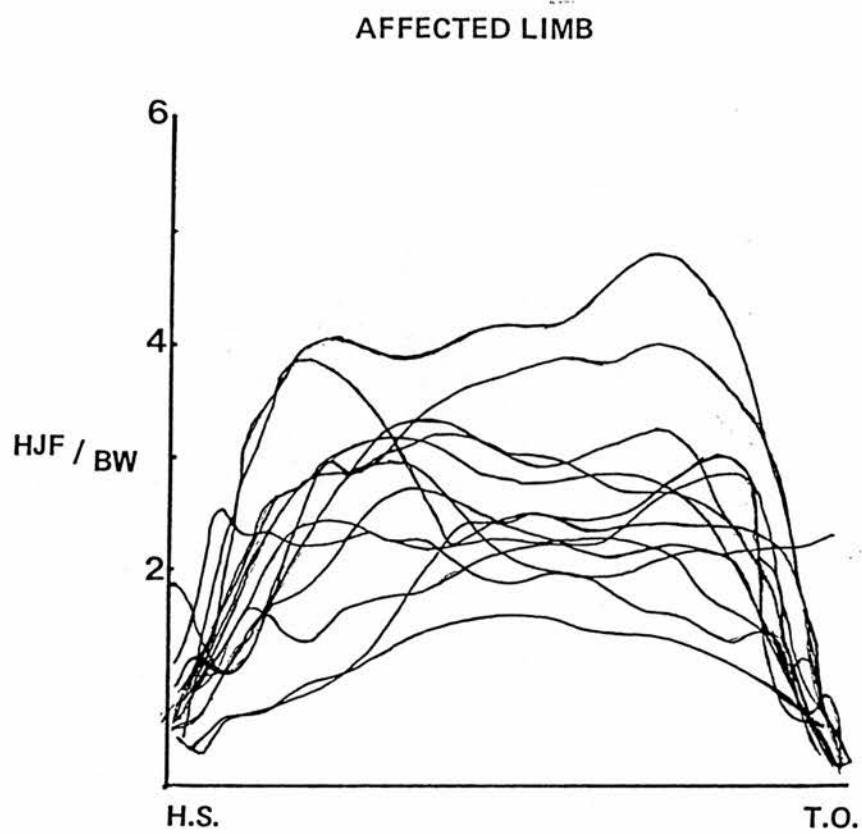


Fig. 4.20 Hip Joint Loads - Affected Hips.

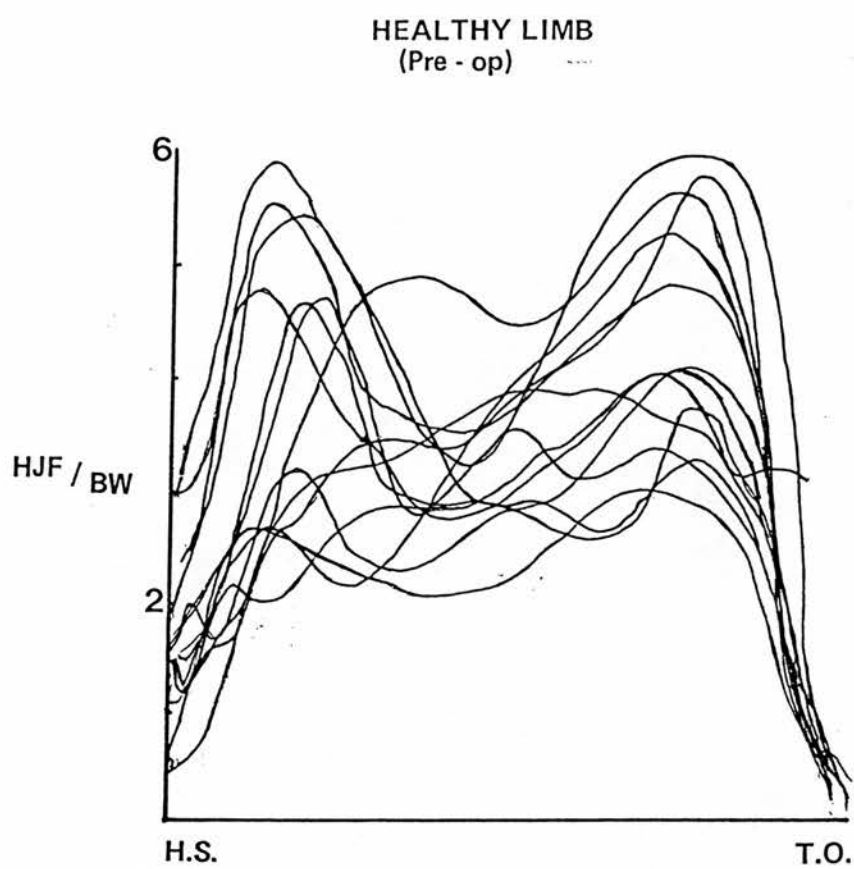


Fig. 4.21 Hip Joint Loads - Healthy Hips.

healthy hips and $2.98 \pm 1.78 \times BW$ for the diseased hips; $p = 0.001$ paired t-test) and most patients show a two peaked curve similar to that described by Paul for normal subjects.

In the diseased hips the majority of patients exhibit a convex curve with poorly defined peaks indicative of abnormal abductor and extensor muscle function in early stance and, to a lesser extent, abductor and flexor muscle function in late stance.

4.III HIP JOINT REPLACEMENT GROUP

The clinical details for these patients are shown in Table 4.1. Note that patient 25 is included in the Girdlestone excision arthroplasty group (Table 4.3).

Although the patients in this study have been treated with two different types of hip arthroplasty the gait parameters for each group are similar and the post-operative test that will be considered in detail is relevant to all patients. The effects of design and anatomical variations will be discussed in the second part of this Chapter.

1. Patient Test

The one year post-operative test for patient 15 will now be presented. This lady, the same as that presented in 4.1, underwent C.A.D. Müller arthroplasty of her right hip.

a. Time-Distance Parameters

Again stride length, stride width and cadence were not measured. The average forward velocity was 1.40m.s^{-1} compared with 0.84 m.s^{-1} pre-operatively (normal group 1.40m.s^{-1}). The duration of stance phase in the operated limb was 0.56 sec. and 0.56 sec. in the healthy limb.

b. Limb Movements

The asymmetry of limb movement seen pre-operatively persists (Figs. 4.22 and 4.23) although, at the hip, ranges have increased on the operated side, particularly in extension. In the healthy limb the flexion and extension ranges at the hip are more like those seen in the normal subject. However, it should be noted that the magnitude

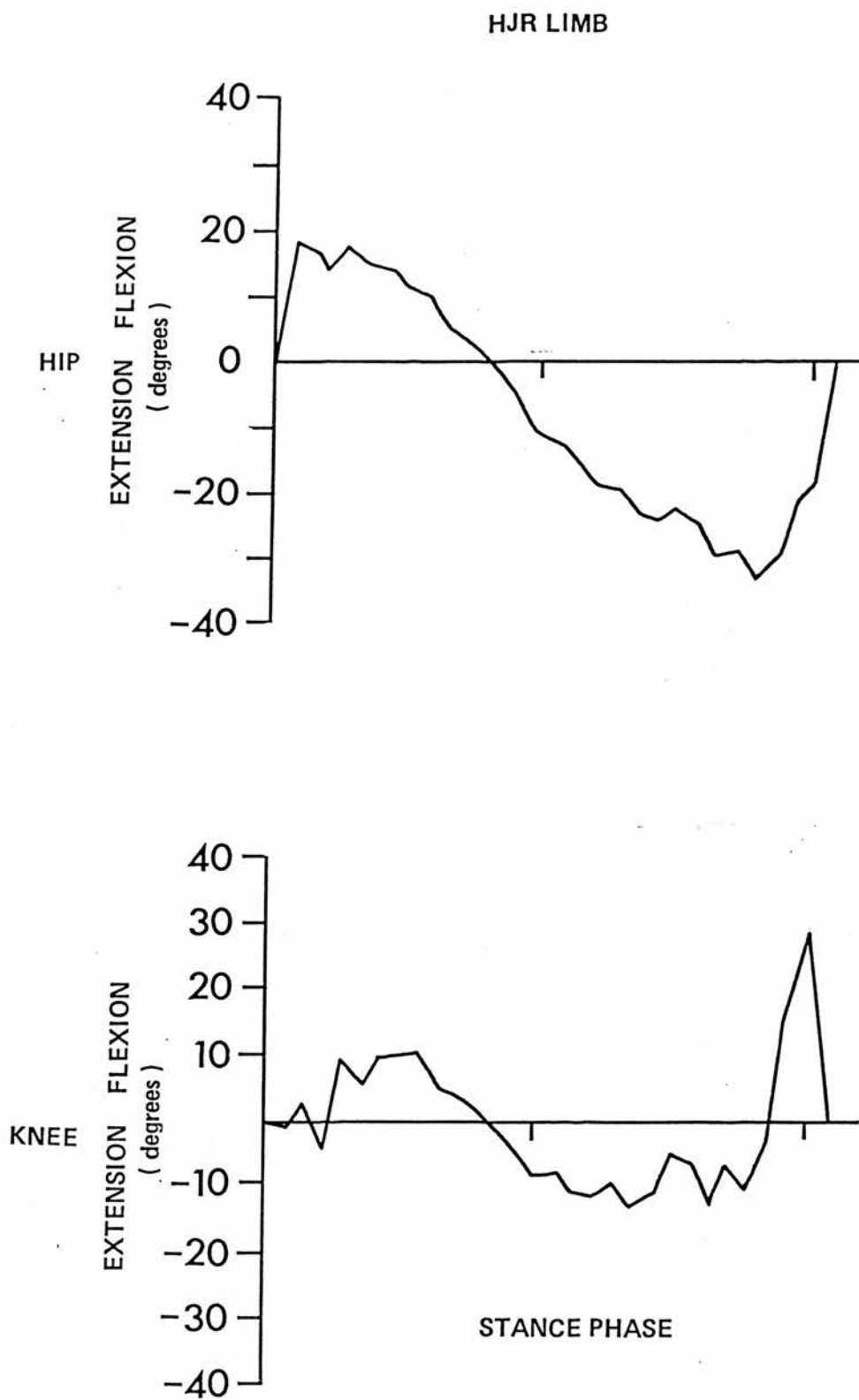


Fig. 4.22 Range of Movement at Hip and Knee Joints
- HJR Limb.

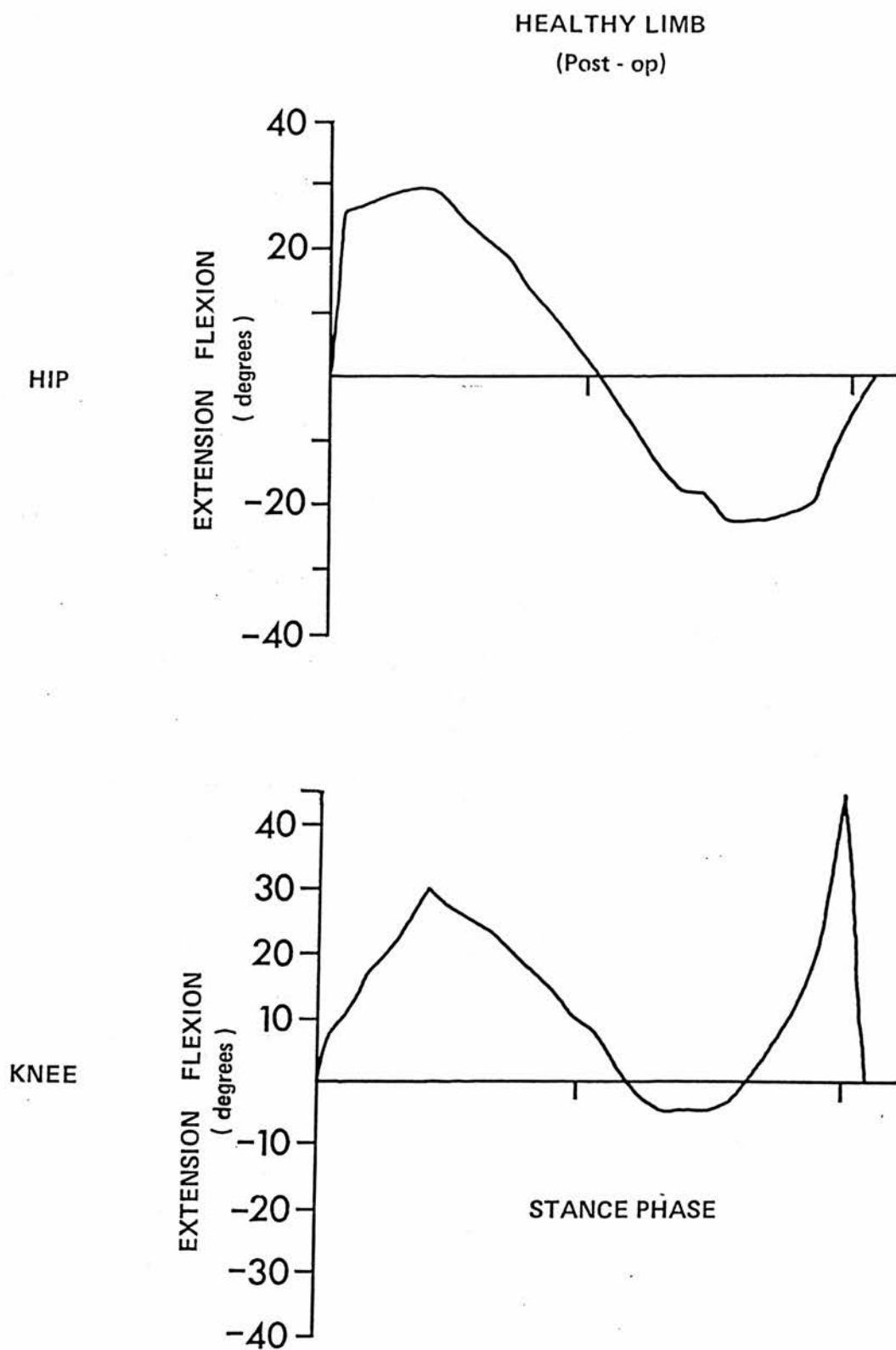


Fig. 4.23 Range of movement at hip and knee joints - healthy limb.

of the excursions into flexion and extension seen in this patient are atypical of the group. Additionally, it was more usual for there to be a greater gain in the flexion range at the replaced hip (vide infra).

The pattern of movement at the healthy knee joint has returned to normal although the initial degree of flexion is large and the knee seems to hyperextend at mid-stance. This feature is also seen at the knee of the operated side where the pattern of movement is very similar to that seen pre-operatively. The hyperextension which is of moderate degree in the patient, is uncommonly seen in the group (vide infra Figs. 4.38 and 4.39) and may represent error in the placing of the knee joint centre markers.

c. Ground-Foot Reaction Forces

There is very much greater symmetry of the floor reaction forces (Figs. 4.24 and 4.25) compared to the pre-operative situation. The fore and aft force (FXG) is symmetrical about mid-stance in both limbs although it is of lesser magnitude in the operated limb. The trough in FYG is not so pronounced on the operated side but the typical pattern has been restored in both limbs, with a more rapid rise to the first peak than pre-operatively. The patterns of FZG remain similar to those seen pre-operatively with a larger laterally directed peak at heel strike and a lateral peak before toe-off on the healthy side. On the operated side the force remains negative in sign, that is, medially directed, for the majority of stance phase, and there is no terminal lateral force.

HJR LIMB

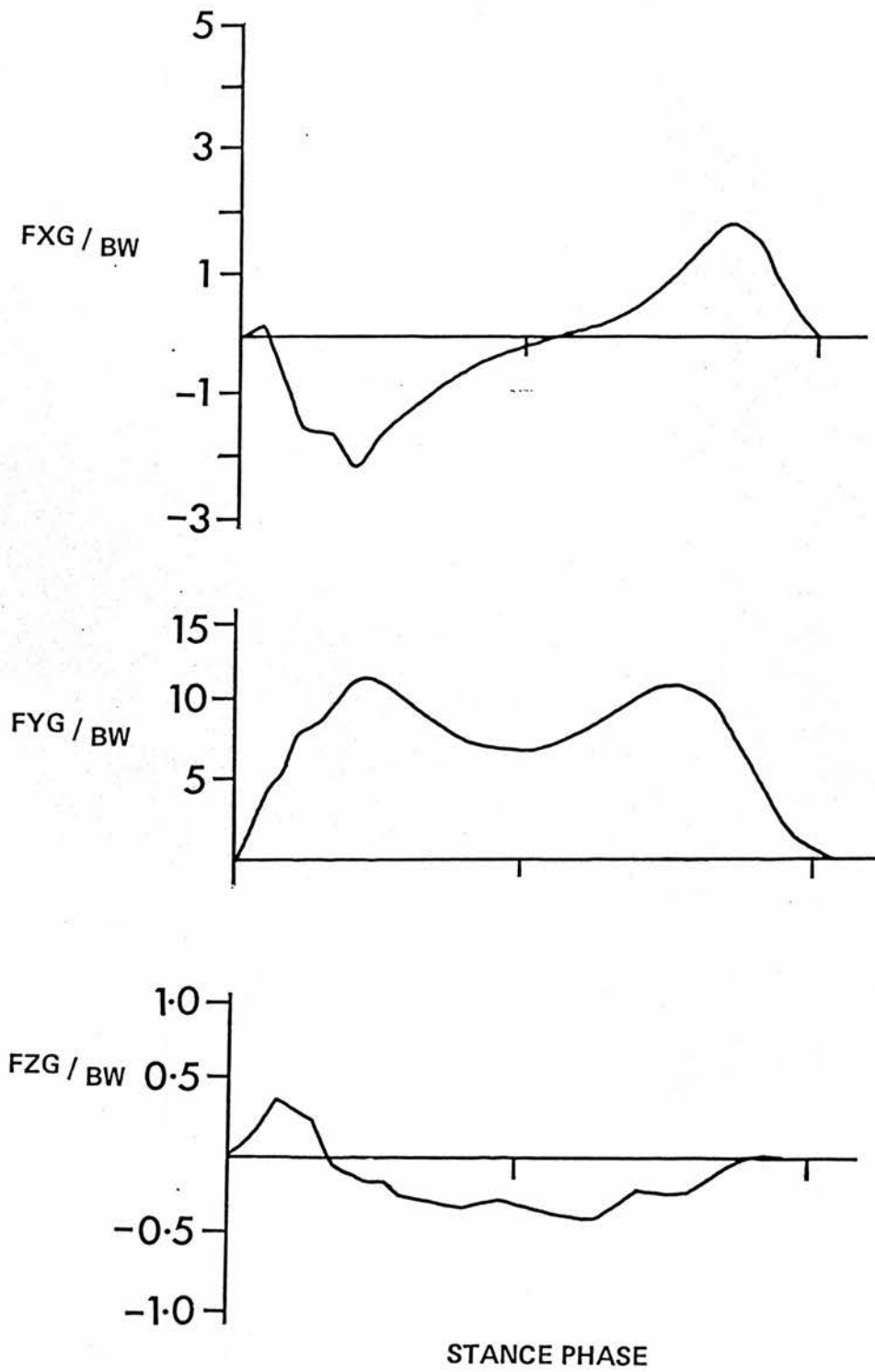


Fig. 4.24 Ground Reaction Forces - HJR limb.

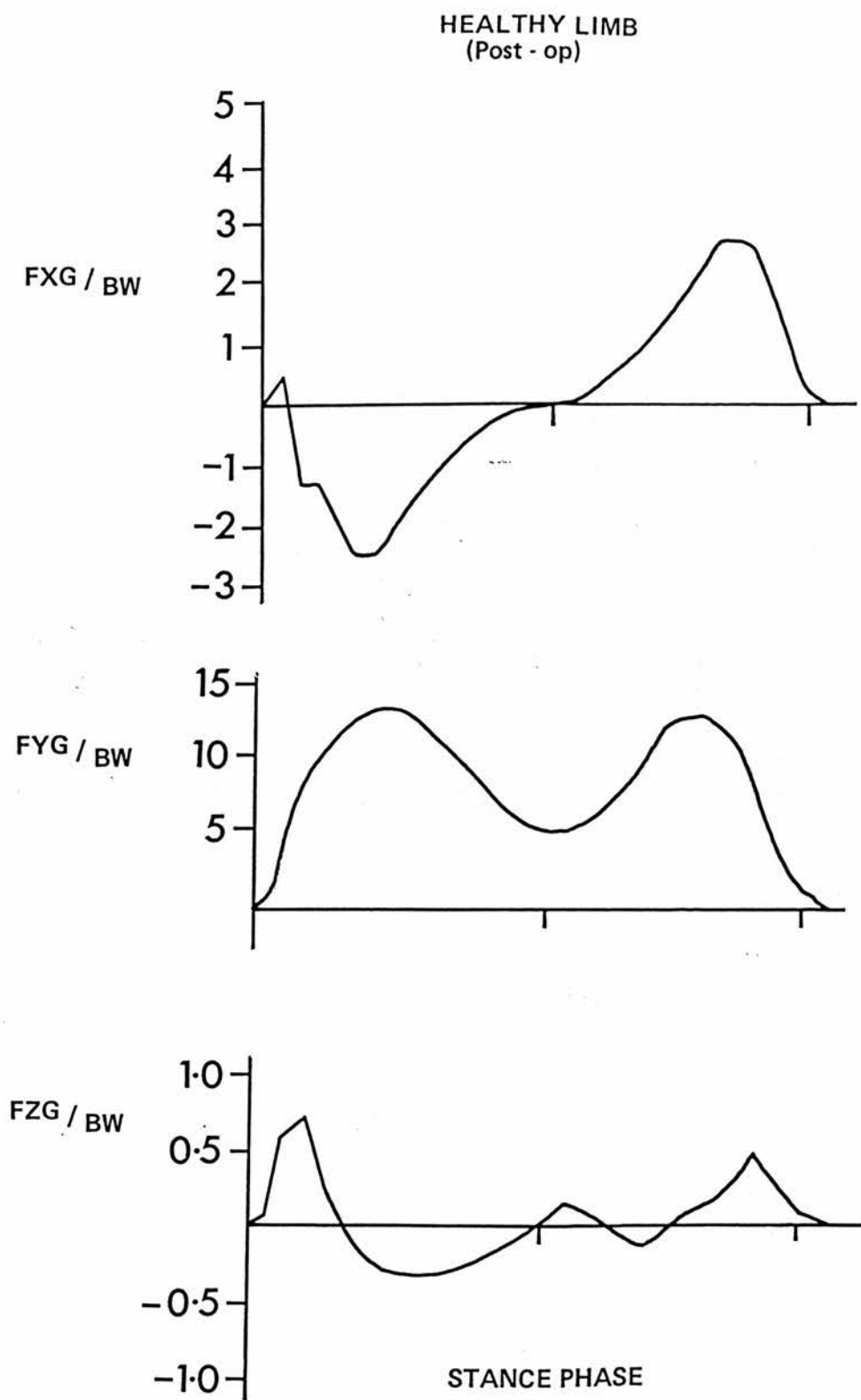


Fig. 4.25 Ground Reaction Forces - healthy limb.

d. Resultant Moments at the Hip and Knee

Hip Joint (Figs. 4.26 and 4.27)

MHX regains its two peak and trough pattern very well on the healthy side but remains abnormal in shape and lower in magnitude on the replaced side indicating the need for less abductor muscle force to balance it than on the healthy side. This return of the pattern on the healthy side is atypical of the group.

MHY remains small in magnitude and indicative of external rotator muscle action on the operated side. However, on the healthy side it has increased in magnitude greatly and reflects the need for both internal and external rotator muscle activity consequent upon greater rotational movement of this limb.

MHZ shows distinct differences between the limbs. The healthy hip requires extensor muscle action in early stance but little in the second part of stance when a small negative deflection will have to be balanced by flexor muscle activity. In the replaced hip, however, the initial positive moment is of short duration, the moment rapidly becoming negative and remaining so. The change from positive to negative precedes the change from hip flexion to extension but results from the combined effect of hip and knee flexion causing the foot to pass below the hip joint early in stance resulting in the line of action of the resultant ground force passing posterior to the hip joint.

HJR LIMB

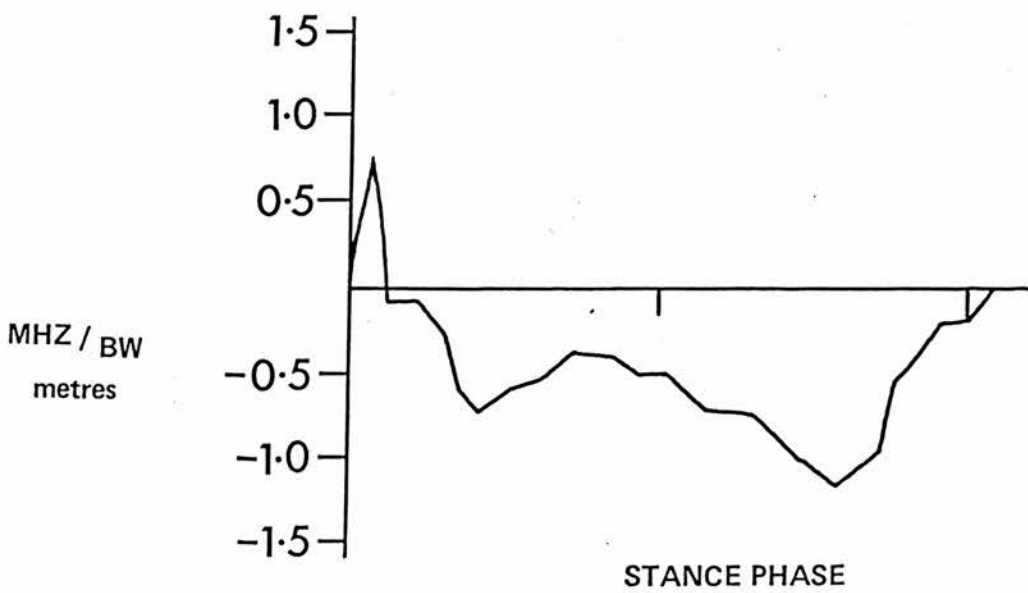
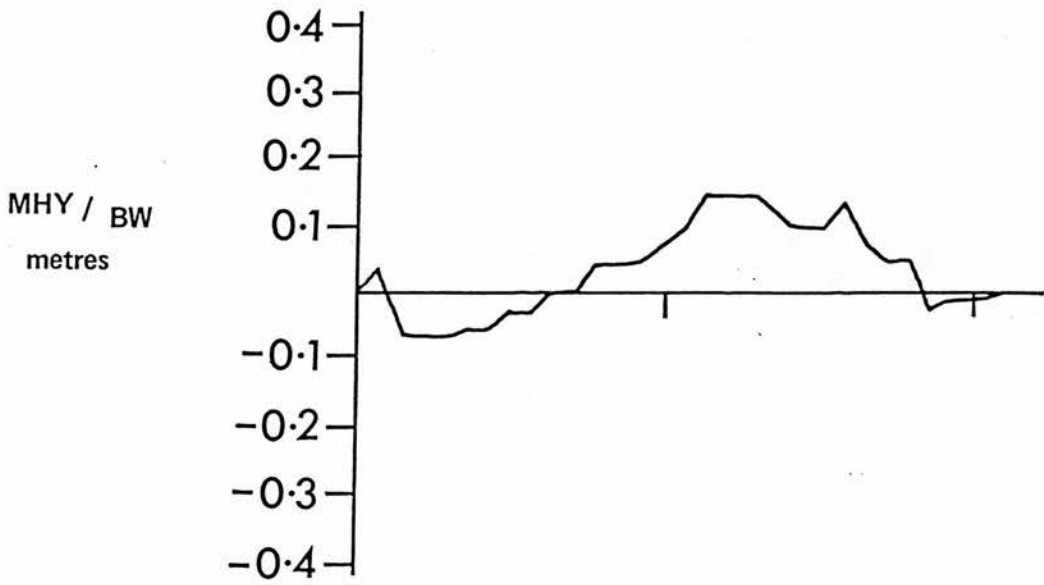
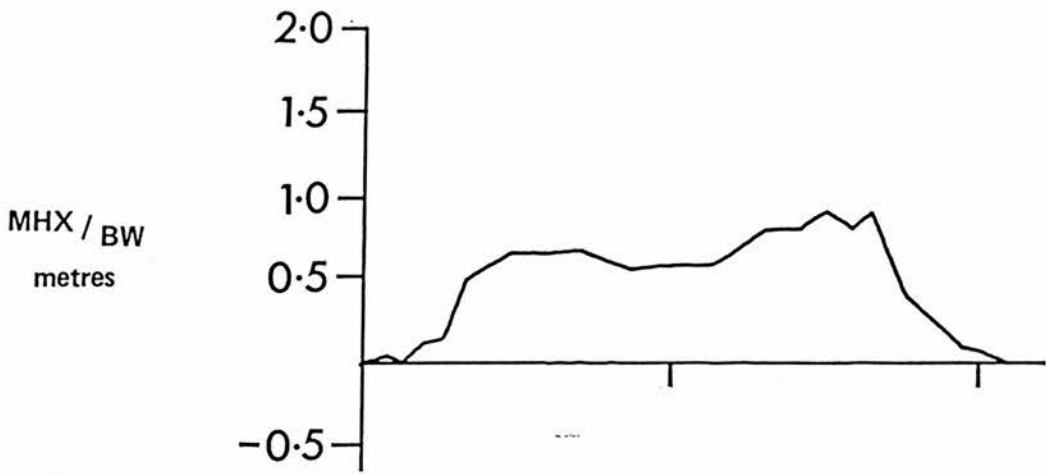


Fig. 4.26 Resultant Moments about the Replaced Hip.

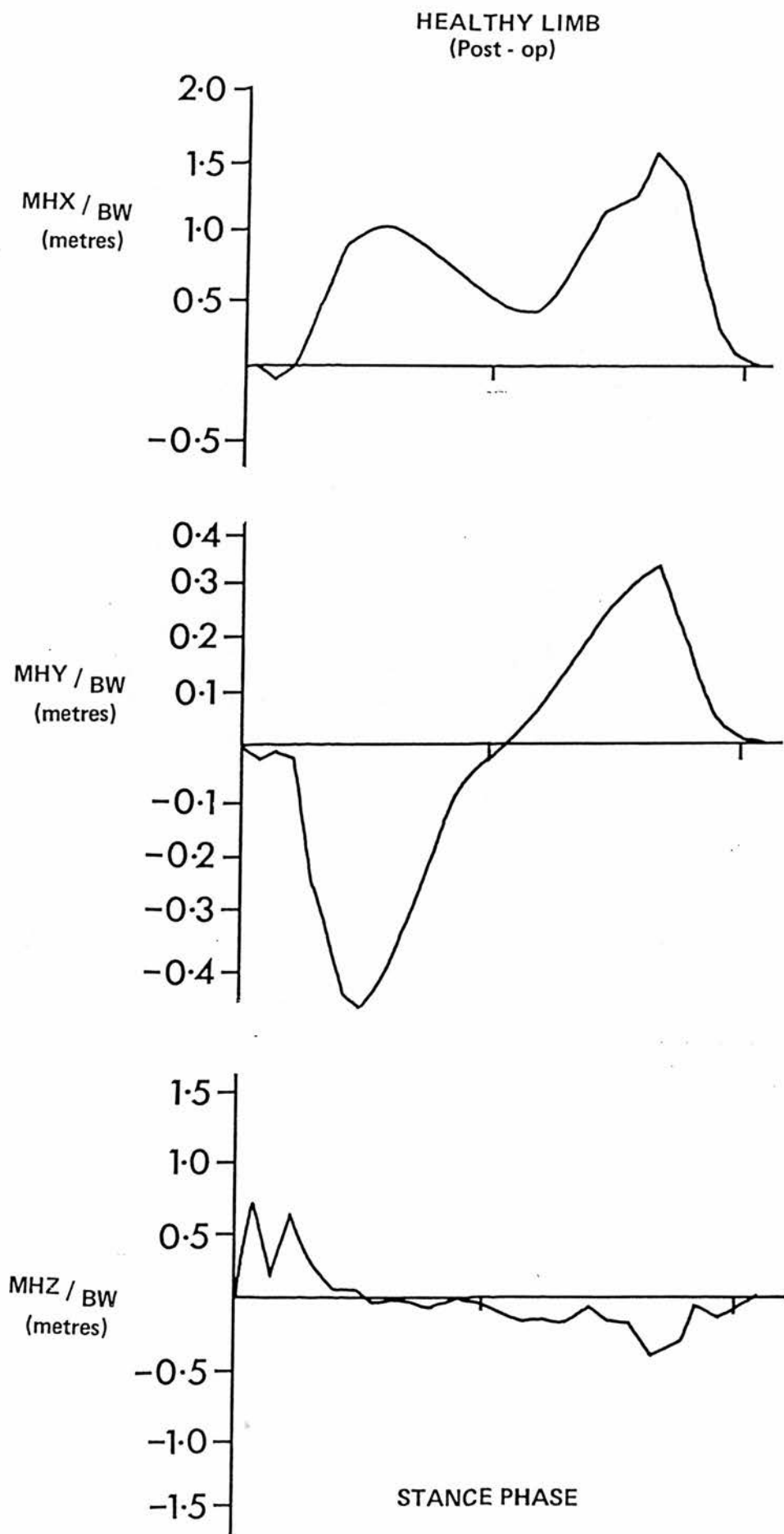


Fig. 4.27 Resultant Moments about the Healthy Hip.

Knee Joint (Figs. 4.28 and 4.29)

The atypical recovery of pattern in the moment about the X-axis (MKX) in the healthy limb is again seen and the failure of recovery of pattern and lower magnitude of moment is seen at the knee of the operated side. This indicates a much more lateral position for the centre of pressure in the latter knee.

MKY shows the same sort of differences as at the hip joint indicating greater rotation of the healthy limb than the operated limb.

MKZ once again reflects the differing pattern of movements at the knees and the positive deflections in the second half of stance, requiring knee flexor muscle activity, correlate with the hyperextension already indicated.

e. Muscle Forces

Knee Joint

The sign of the knee flexor moment (MHZ) indicates the pattern of muscular activity. The pattern in both limbs (Figs. 4.28 and 4.29) differs from that seen pre-operatively and both limbs show a similar pattern, although different magnitudes at one year.

In the first half of stance phase the deflection is largely negative in both limbs indicating a need for activity in quadriceps femoris as seen in Figure 4.30. Quadriceps activity is again necessary in the terminal phase of stance during the period of double support. This differs markedly from the pattern seen pre-operatively (Fig. 4.5) and more closely resembles the normal subject.

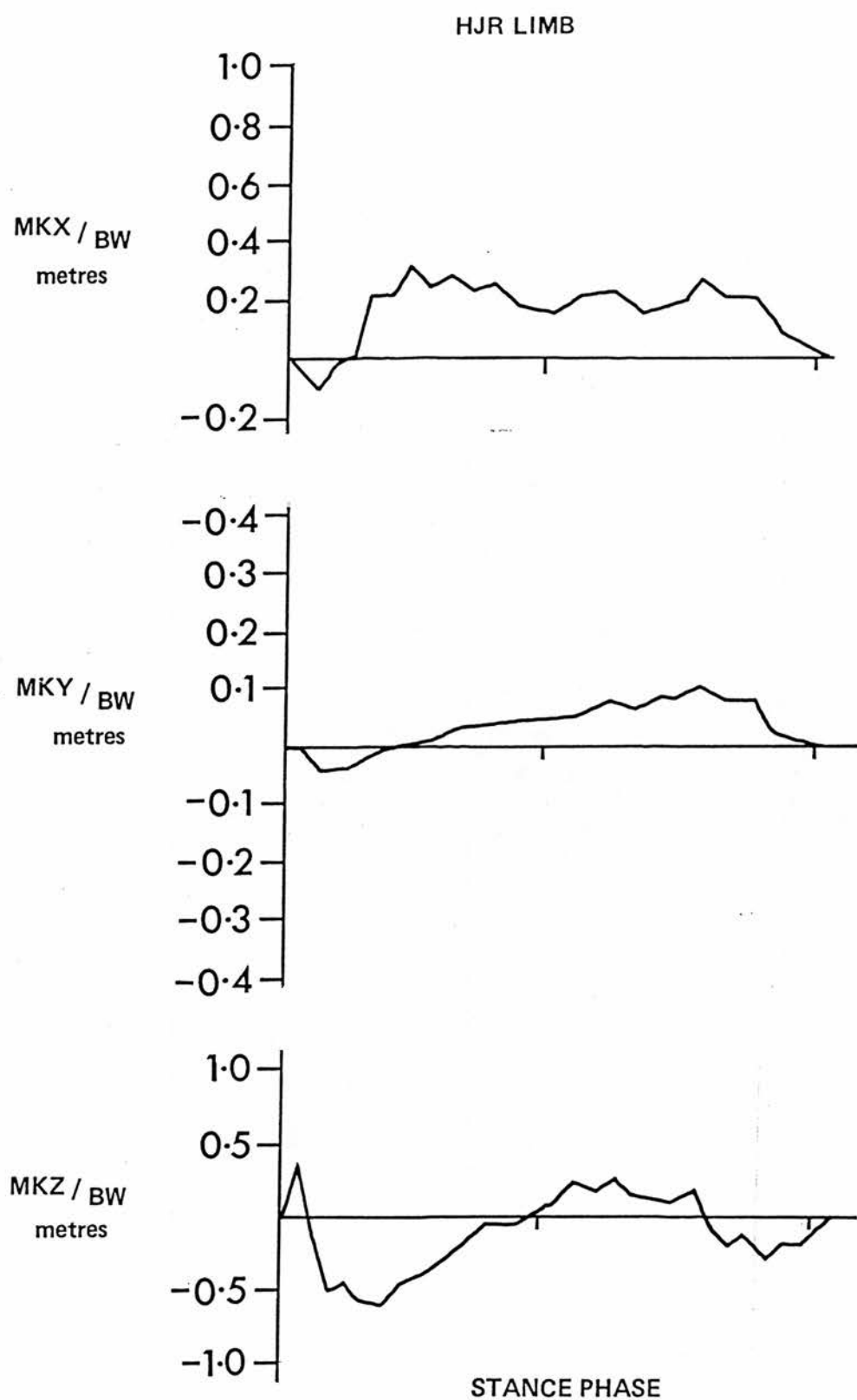


Fig. 4.28 Resultant Moments about the Ipsilateral Knee.

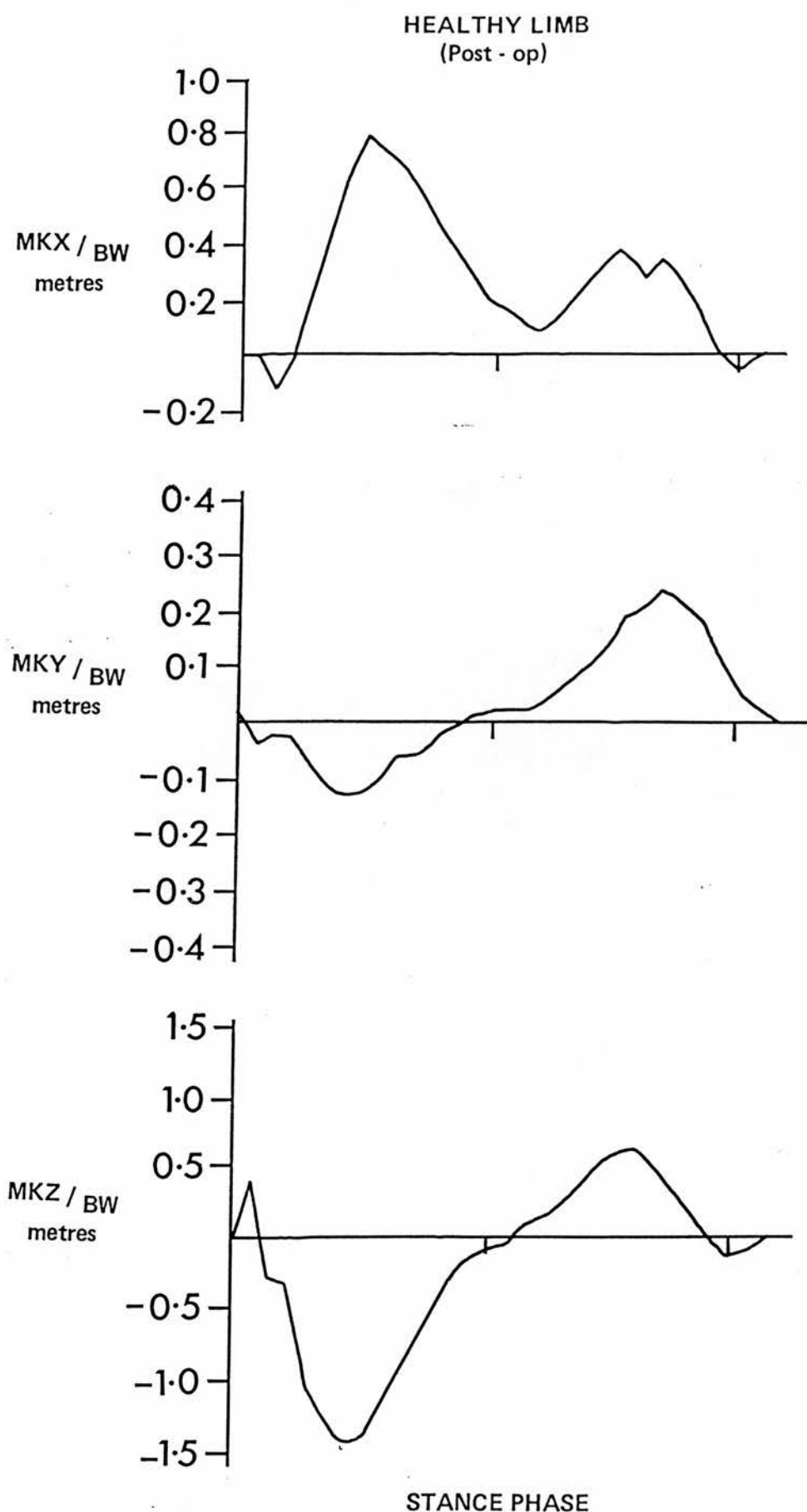
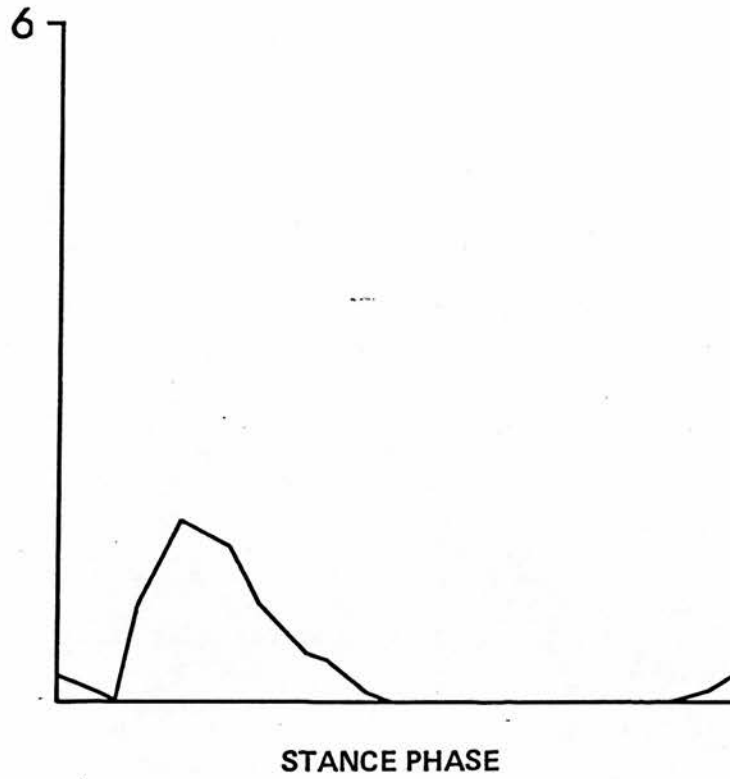


Fig. 4.29 Resultant Moments about the Contralateral Knee.

HJR LIMB

QUADRICEPS
MUSCLE FORCE / BW



HEALTHY LIMB
(Post - op)

QUADRICEPS
MUSCLE FORCE / BW

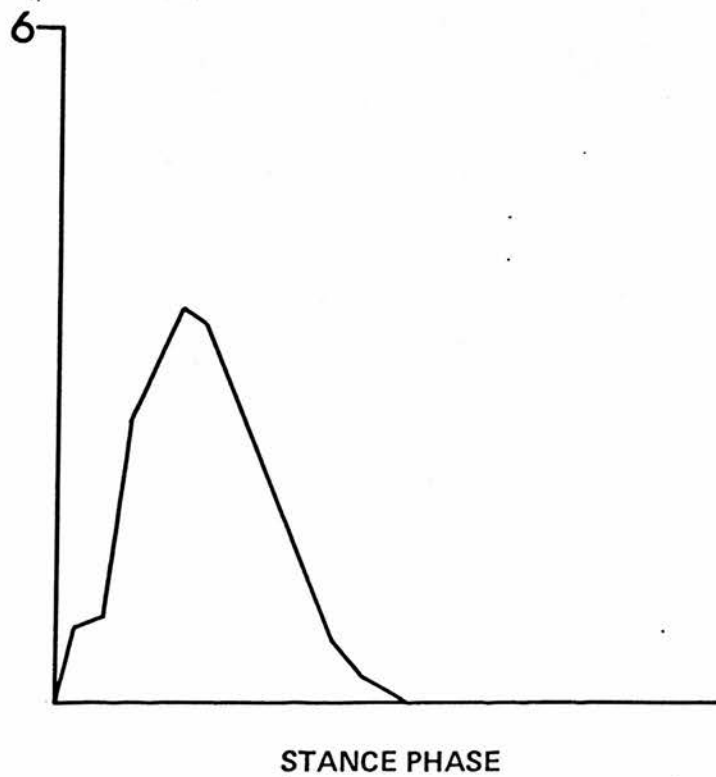


Fig. 4.30 Quadriceps Muscle Force.

Immediately following heel strike and in the second half of stance phase the deflection in MHZ is positive in both limbs. After heel strike the flexor-extensor moment at the hip joint is also positive, and, as in the pre-operative situation, the magnitude of the moment at the hip exceeds that at the knee resulting in hamstrings force being allocated to balance MKZ. In the second half of stance, however, MHZ is negative and hence MKZ will be balanced by activity in gastrocnemius.

Hip Joint

Inspection of the resultant moments about the hip provides a guide to the pattern of muscular activity.

The difference in magnitude and pattern of MHX (Figs. 4.27 and 4.28) is manifest in a different pattern and magnitude of abductor muscle force between the hips (Fig. 4.31). This is particularly notable in early stance.

The differences in MHZ have already been discussed (vide supra), and the differences in flexor-extensor muscle action can be seen in Figures 4.32 and 4.33. The prolonged action of hip flexor muscles on the side of the replaced hip is of particular note and reflects the pattern of movement seen at the hip and knee joint of this limb (Fig. 4.22).

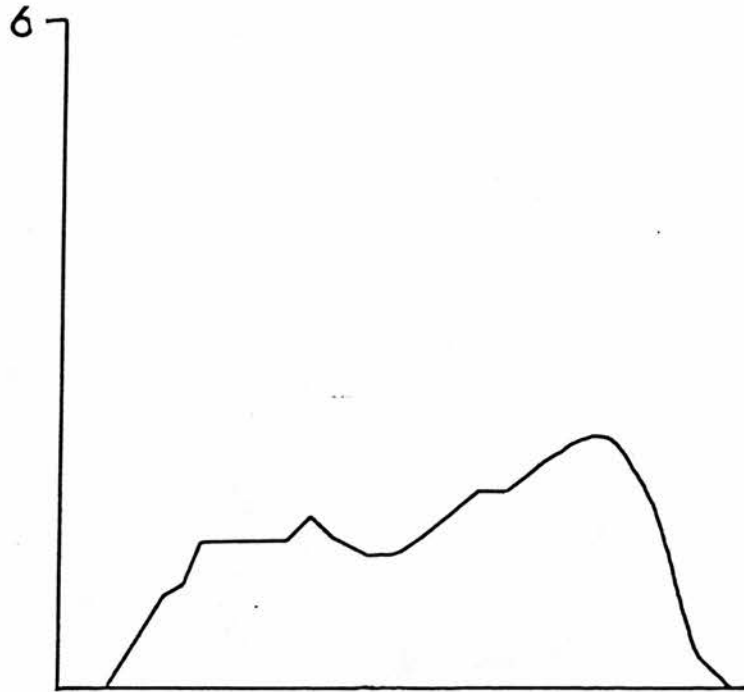
f. Joint Loads

Knee Joint

The total knee joint load for both limbs is seen in Figure 4.34. Unlike the pre-operative curves these do not

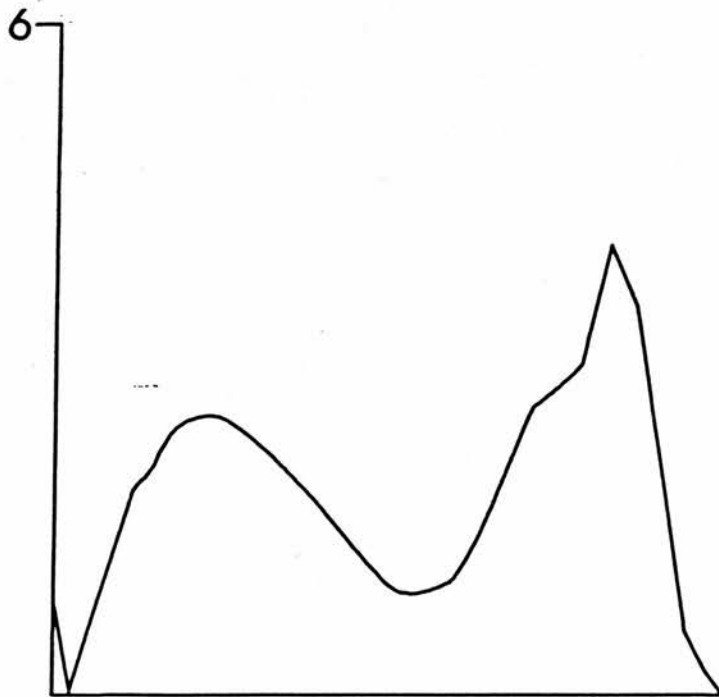
HJR LIMB

ABDUCTOR
MUSCLE FORCE / BW



HEALTHY LIMB (Post - op)

ABDUCTOR
MUSCLE FORCE / BW

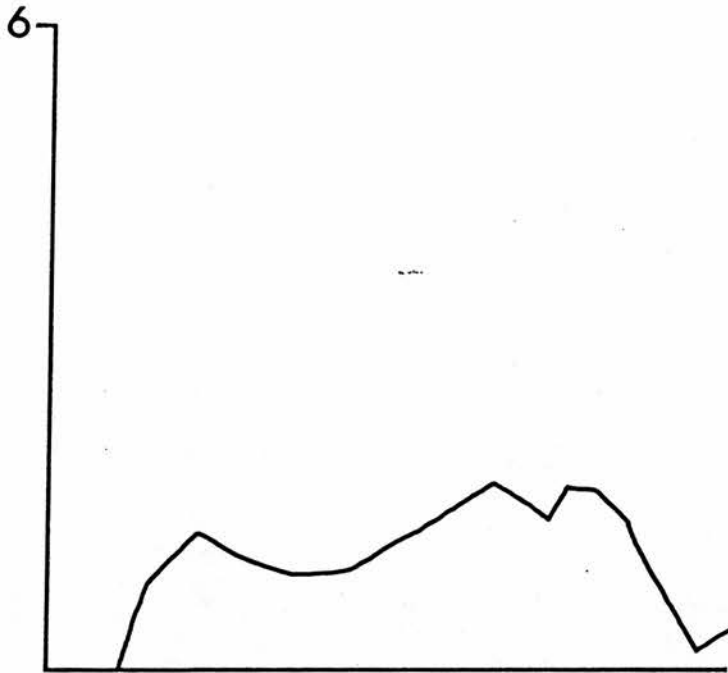


STANCE PHASE

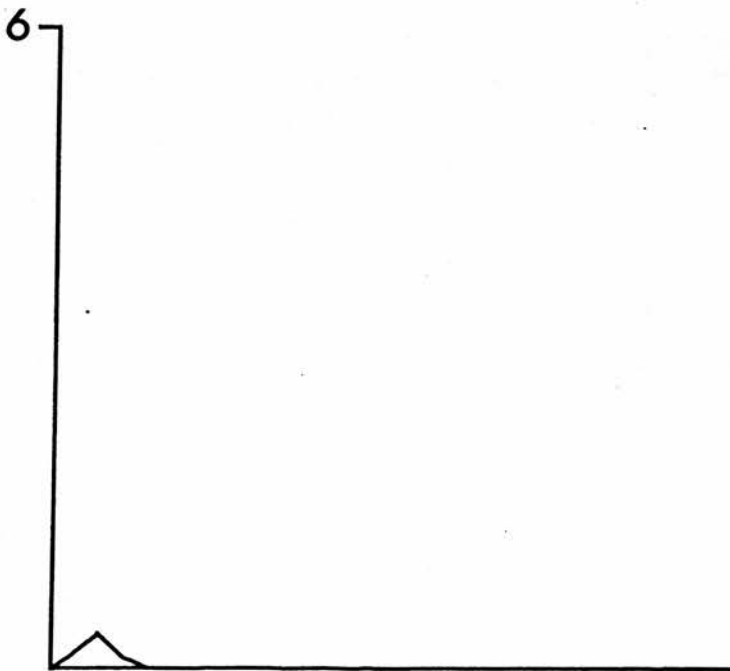
Fig. 4.31 Hip Abductor Muscle Force.

HJR LIMB

HIP
FLEXOR MUSCLE
FORCE / BW



TOTAL
EXTENSOR MUSCLE
FORCE / BW

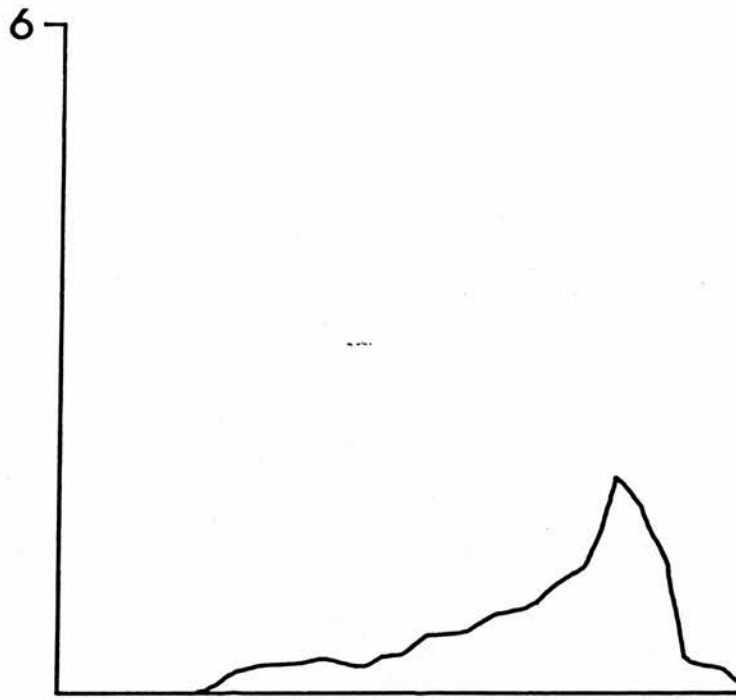


STANCE PHASE

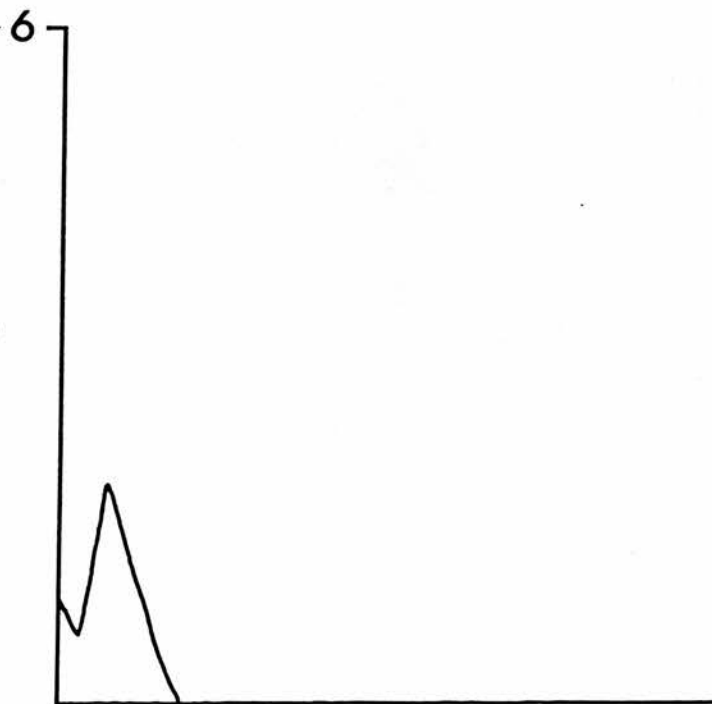
Fig. 4.32 Hip Flexor and Extensor Muscle Force - HJR Limb.

HEALTHY LIMB
(Post - op)

HIP
FLEXOR MUSCLE
FORCE / BW



TOTAL
EXTENSOR MUSCLE
FORCE / BW



STANCE PHASE

Fig. 4.33 Hip Flexor and Extensor Muscle Force - Healthy Limb.

resemble the curves for quadriceps femoris. The magnitude of the joint force is greater on the healthy side in early stance reflecting the greater degree of flexion and greater quadriceps femoris force. However, the magnitude of the forces in the second half of stance phase, due to gastrocnemius action, are very similar.

The pattern for both limbs more closely resembles that for the normal subjects but the initial peak (Morrison's 'b' peak) is of greater magnitude than normal in the healthy limb (4.5 xBW). Unlike the pre-operative situation, three peaks cannot be discerned, the 'a' peak being absent. Since this results from hamstrings action it might be expected to be absent in the operated limb where activity in this muscle is small and short-lived. Indeed, the rise to the first peak of knee joint force is more gradual in this limb than in the healthy limb. In the latter limb there is an initial higher force and a more rapid rise to peak force. It may be that the brief 'a' peak has been "missed" by the rate of data sampling used.

The compartmental distribution of the joint load is shown in Figures 4.35 and 4.36. Both limbs transmit most of the load through the medial compartment of the joint although at the maxima the lateral compartments share some of the load.

Hip Joint

The total hip joint loads, shown in Figure 4.37, are very similar in magnitude for each limb, but the pattern differs, the trough being much more pronounced on the healthy side. The replaced hip therefore carries load at

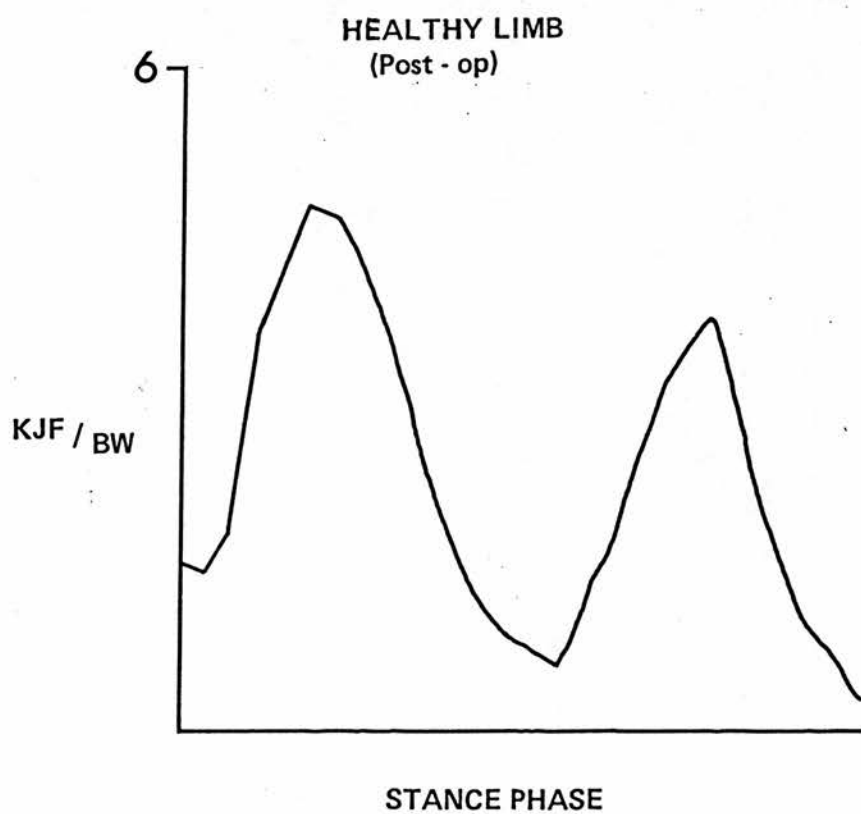
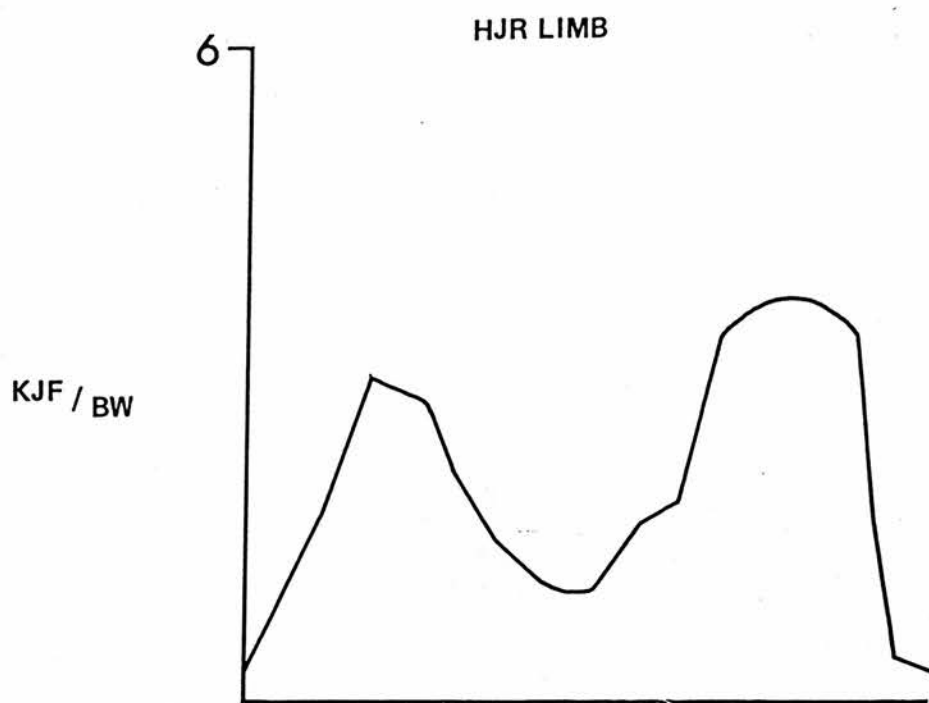
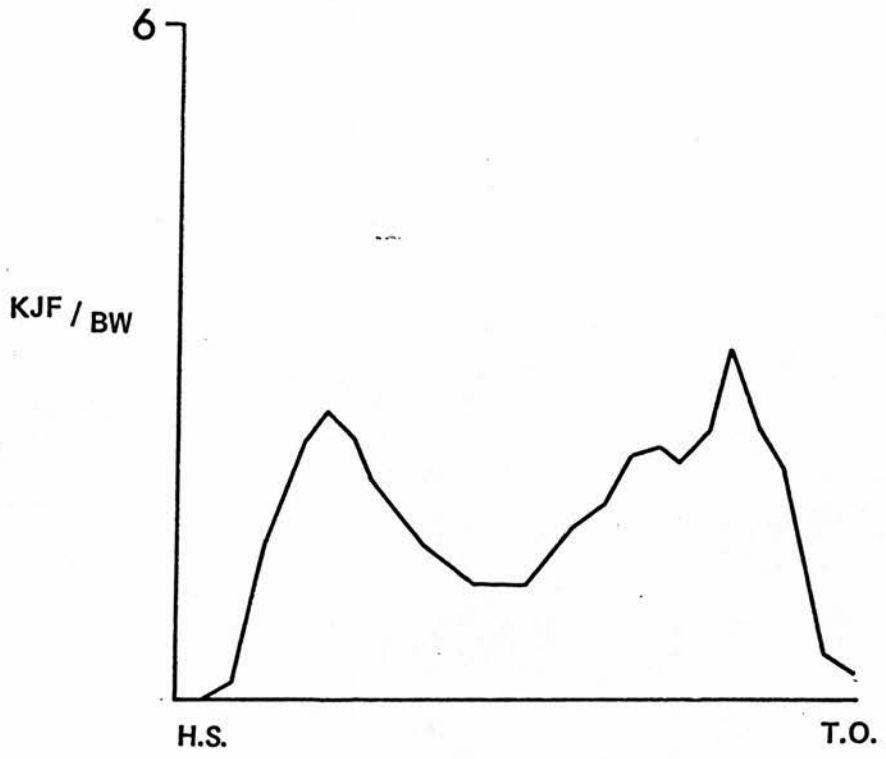


Fig. 4.34 Knee Joint Load.

HJR LIMB

LATERAL
JOINT COMPARTMENT



MEDIAL
JOINT COMPARTMENT

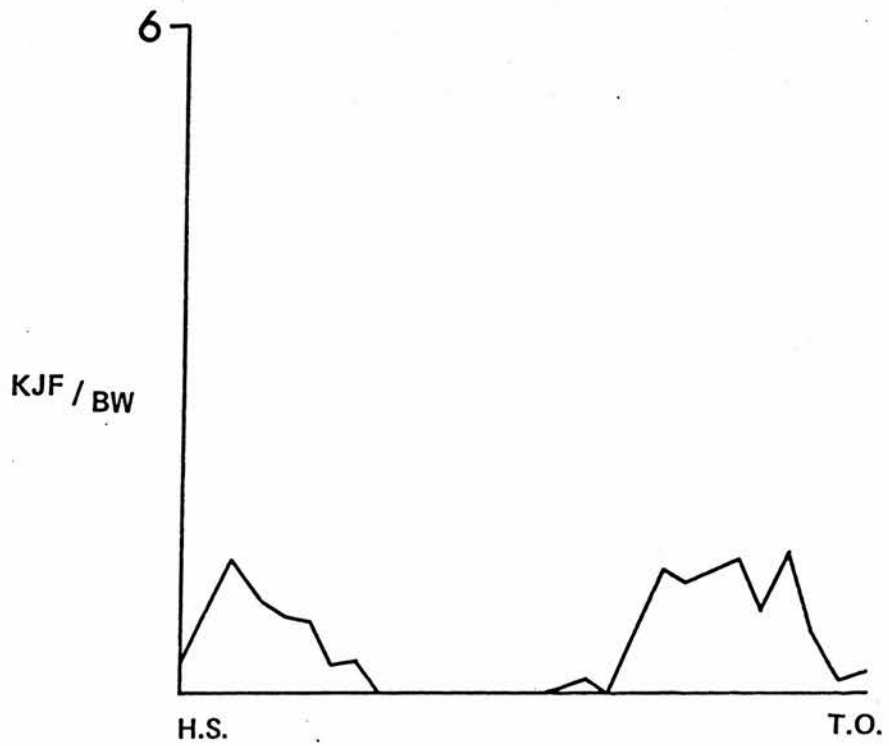


Fig. 4.35 Compartmental Knee Joint Load - HJR Limb.

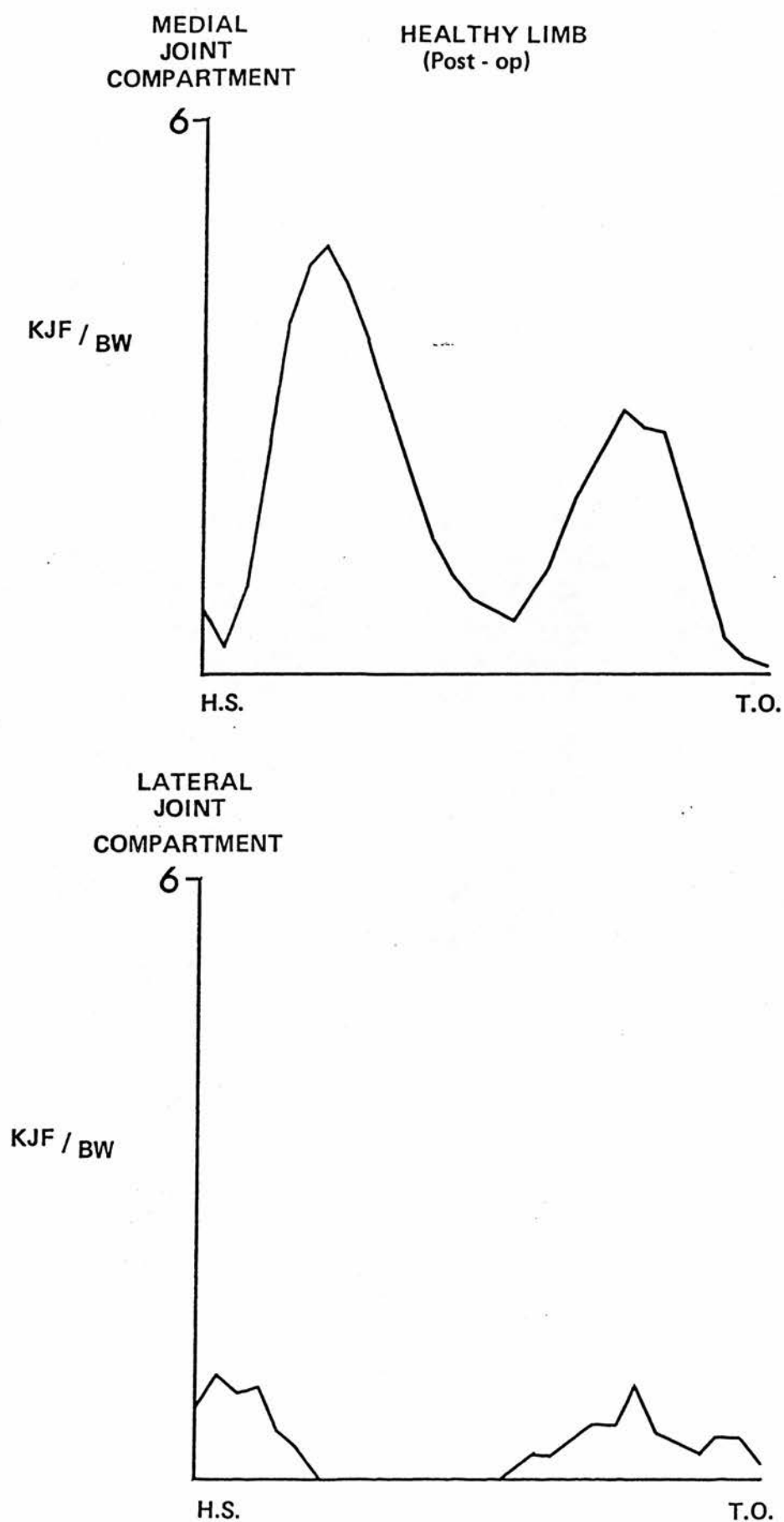
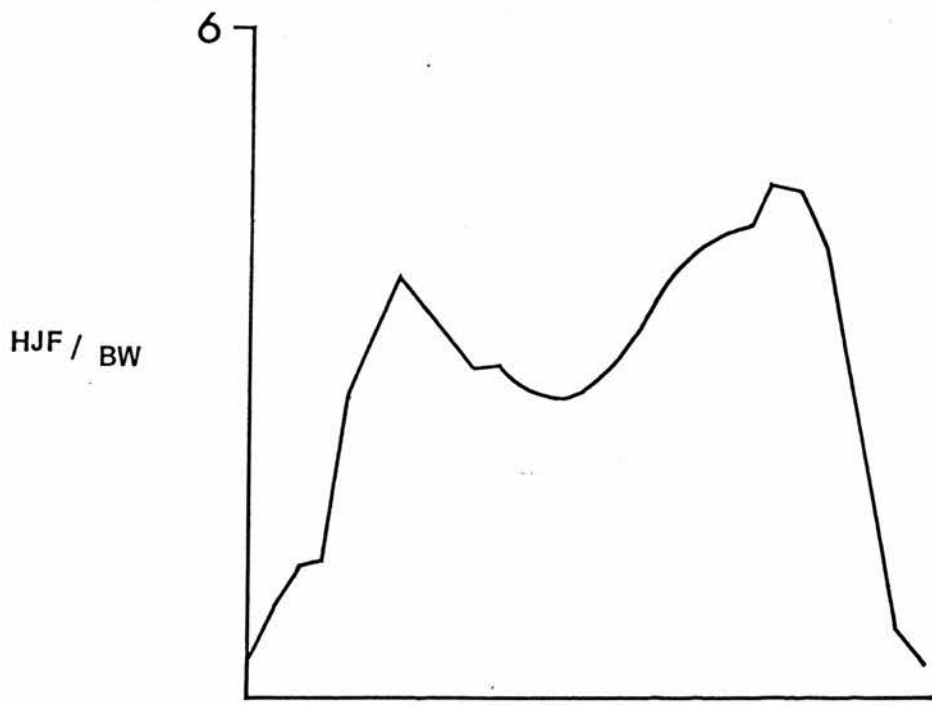
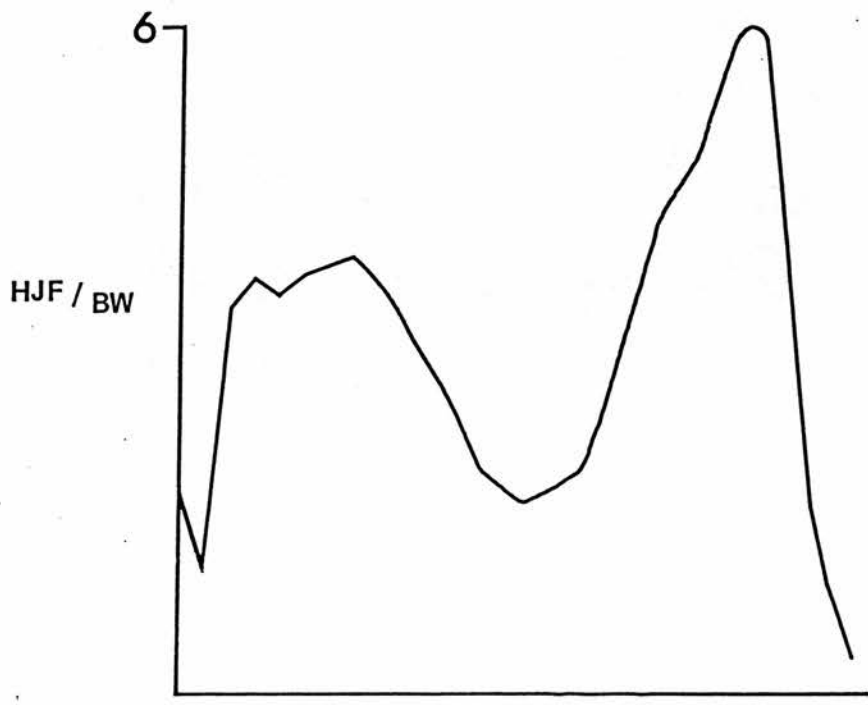


Fig. 4.36 Compartmental Knee Joint Load - Healthy Limb.

HJR LIMB



HEALTHY LIMB
(Post op)



STANCE PHASE

Fig. 4.37 Hip Joint Load

a sustained level throughout stance phase.

The similarity in the magnitude of the joint load in the initial half of stance phase, despite the difference in the magnitude of the abductor muscle force at this phase of gait, is accounted for by the activity in the hip flexor muscles at the replaced joint. This pattern of muscle action was noted pre-operatively.

In the second half of stance the slightly greater magnitude of load at the healthy hip results from the magnitude of the force exerted by the abductor muscles at this joint, the abductor muscles clearly accounting for most of the load across the healthy hip joint.

4.III

2. Summary of Results for Hip Joint Replacement Group

a. Time-Distance Parameters

The mean velocity for this group of patients was $1.17 \pm 0.40 \text{ m.s}^{-1}$, faster than the pre-operative group but slower than the group of young normal subjects. Stance phase remained shorter for the operated limb but the difference between the limbs was less (0.70 ± 0.13 sec. for the healthy limb and 0.68 ± 0.13 sec. for the operated limb).

b. Limb Movements

The asymmetry of movement seen pre-operatively improves greatly post-operatively but certain differences remain (Figs. 4.38 and 4.39).

At the hip joints the major difference is in the variability of the timing of the change from flexion to extension at the replaced hip joint.

At the knee joints, the typical movement pattern is seen in nearly all contralateral knees, but the magnitude of the initial flexion peak is usually greater than that seen in normal subjects (mean $23.94 \pm 16^\circ$; other limb, mean $15.12 \pm 15.4^\circ$). At the ipsilateral knee joints there is an increase in the magnitude of the initial flexion peak when compared to the pre-operative results (Fig. 4.10) but the pattern has changed little.

c. Ground-Foot Reaction Forces (Figs. 4.40 and 4.41)

The curves FXG and FYG are very similar for each limb. FYG shows a rather better developed two peak and trough format on the healthy side but the rate of vertical loading

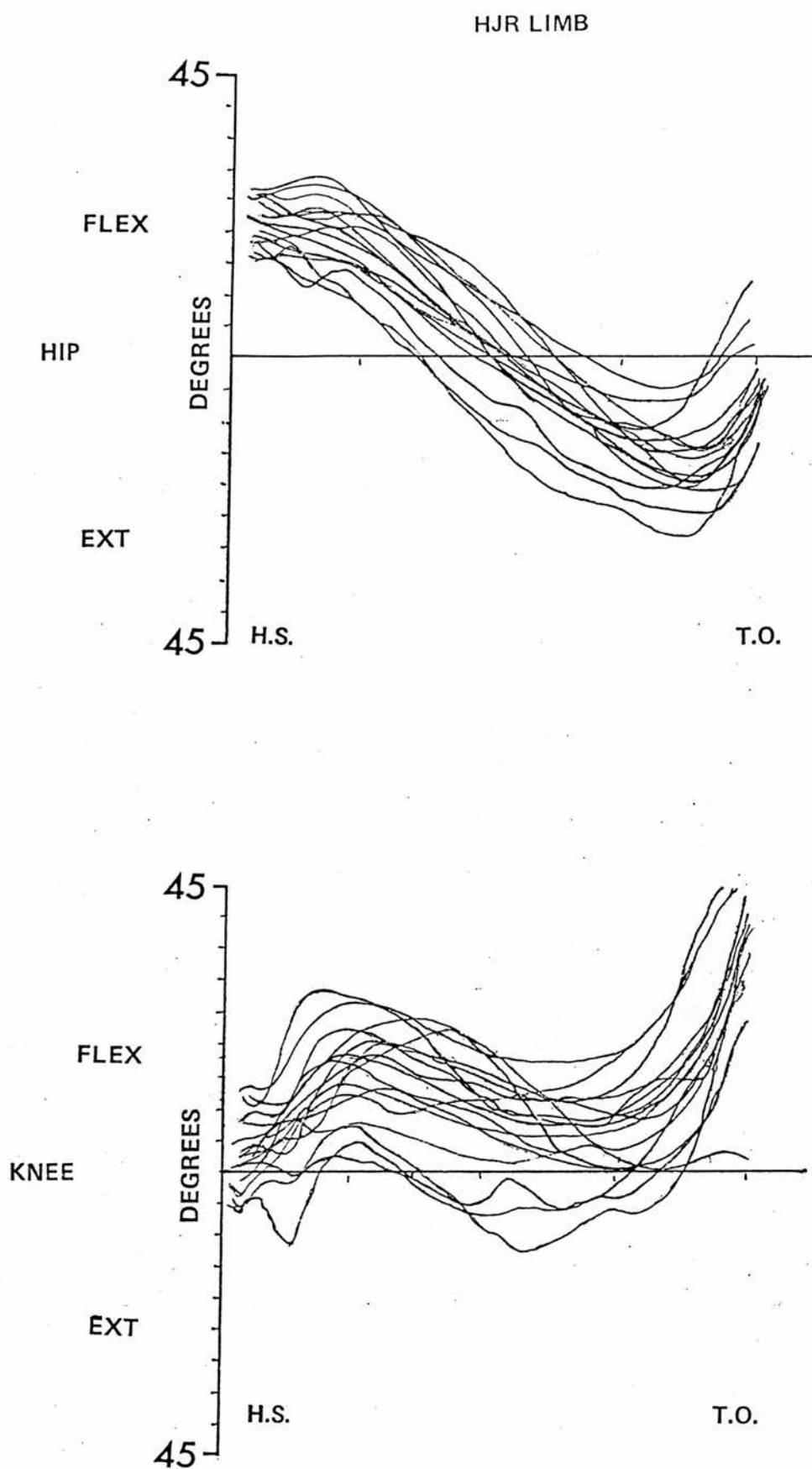


Fig. 4.38 Range of Movement at Hip and Knee Joints
- HJR Limbs.

HEALTHY LIMB
(Post - op)

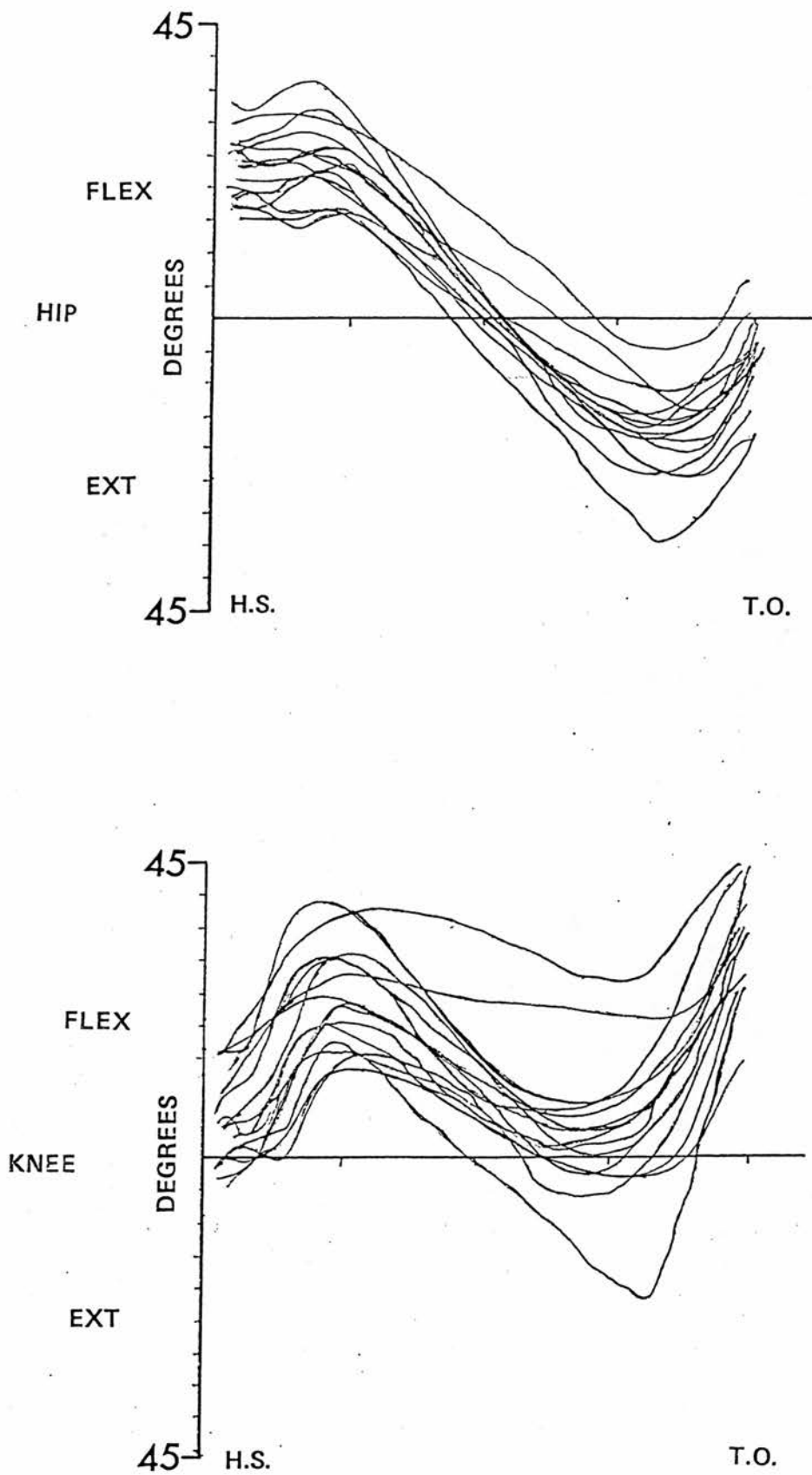


Fig. 4.39 Range of Movement at Hip and Knee Joints
- Healthy Limbs.

HJR LIMB

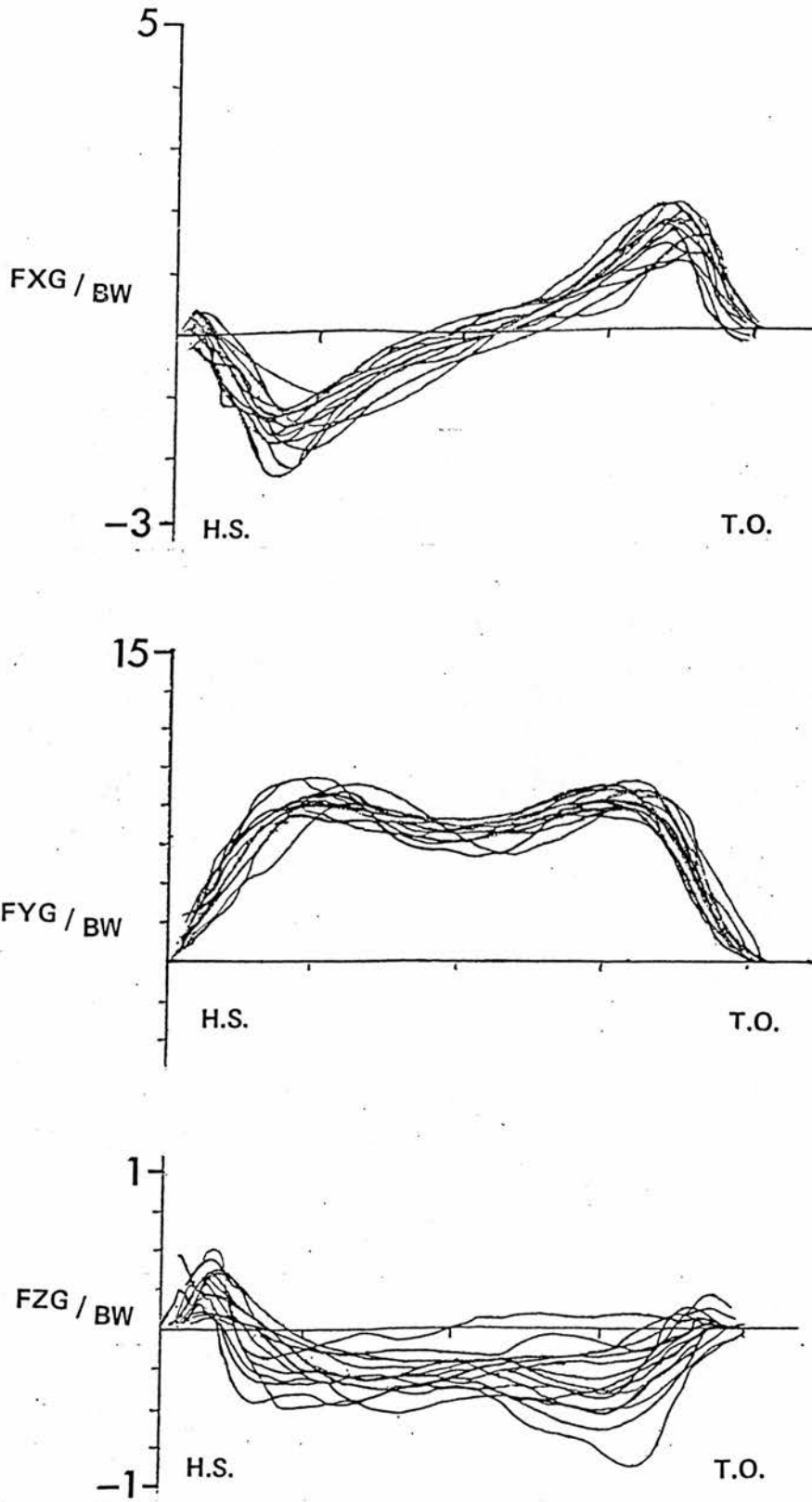


Fig. 4.40 Ground Reaction Forces - HJR Limbs.

HEALTHY LIMB
(Post - op)

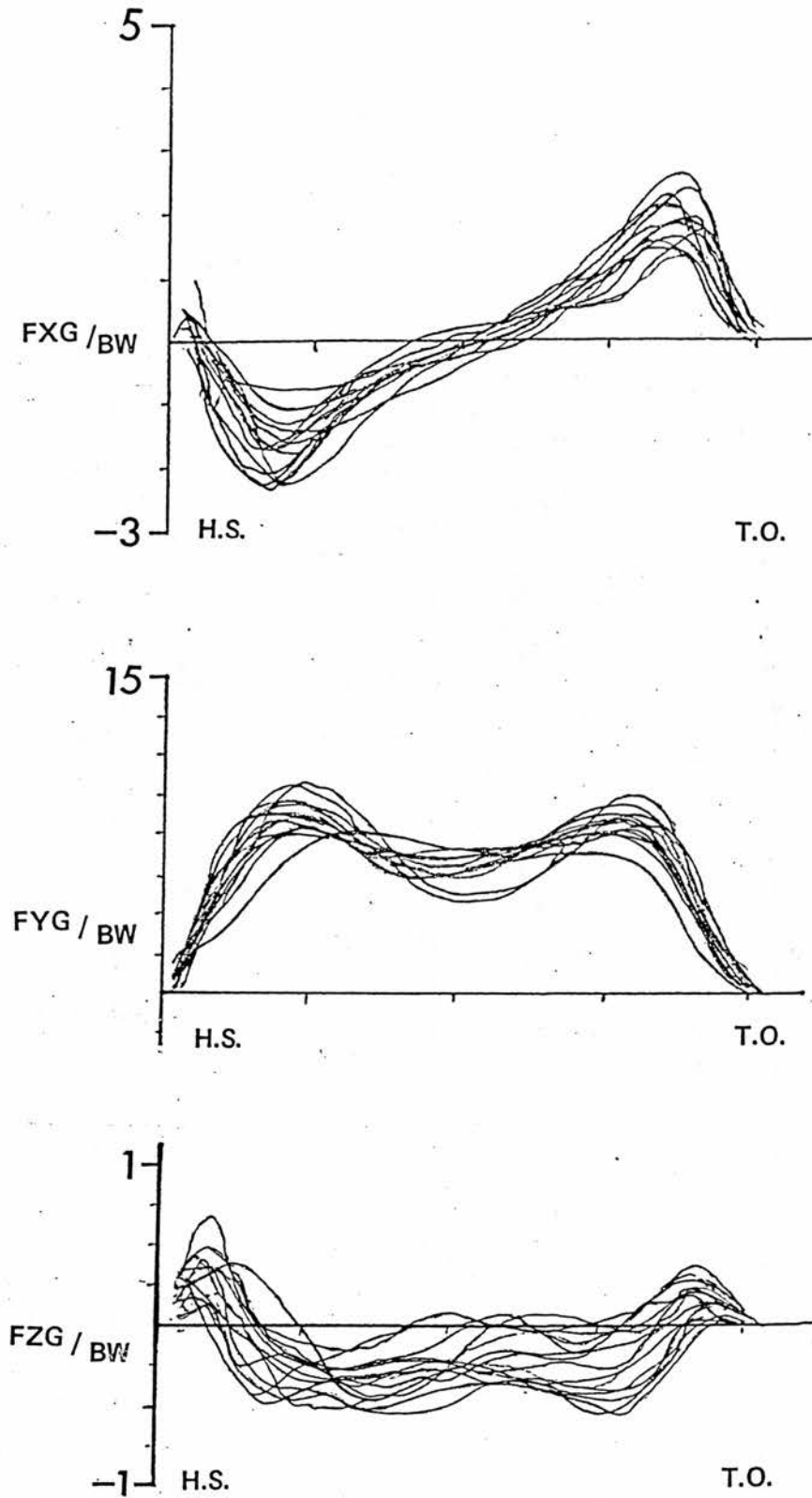


Fig. 4.41 Ground Reaction Forces - Healthy Limbs.

seemingly reduced pre-operatively, appears to have become the same for each limb.

The curves for FZG show less scatter than pre-operatively but maintain the same pattern.

d. Resultant Moments at the Hip and Knee

Hip Joint (Figs. 4.42 and 4.43)

The major difference between the limbs is in the shape and magnitude of the curves for MHX, the abduction-adduction moment. Whilst the healthy hips regain the typical two peak and trough pattern of the normal subject, the replaced hips show irregular plateaux with poorly defined peaks. The magnitude of MHX is also reduced on the side of the replaced hip. MHZ shows the same wide scatter as seen pre-operatively and in normal subjects and no change in pattern is seen.

Knee Joint (Figs. 4.44 and 4.45)

The abduction-adduction moment about the knee joint (MKX) shows a strong tendency towards recovery of the normal pattern in most of the healthy contralateral knees whereas the majority of the ipsilateral replaced limb knees show irregular plateaux as is seen with MHX (Figs. 4.42 and 4.43). The variability in the magnitude of this moment is less for both limbs and the magnitude is usually greater on the healthy side (0.54 ± 0.46 metres for the healthy limb and 0.38 ± 0.32 metres for the ipsilateral knee).

MKZ, the flexor-extensor moment about the knees changes little in configuration or magnitude at the

HJR LIMB

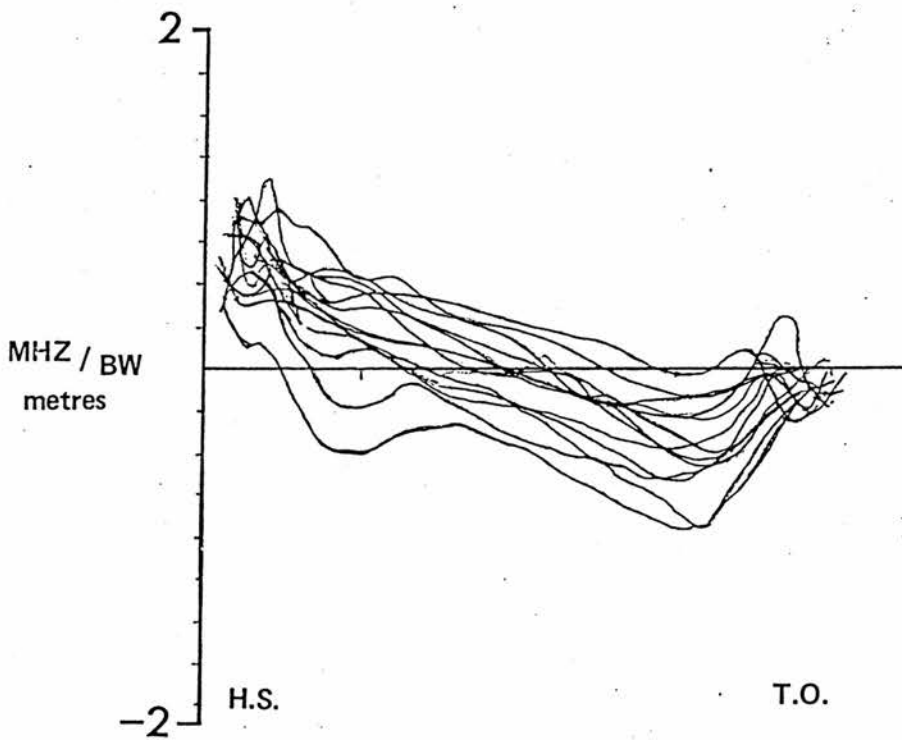
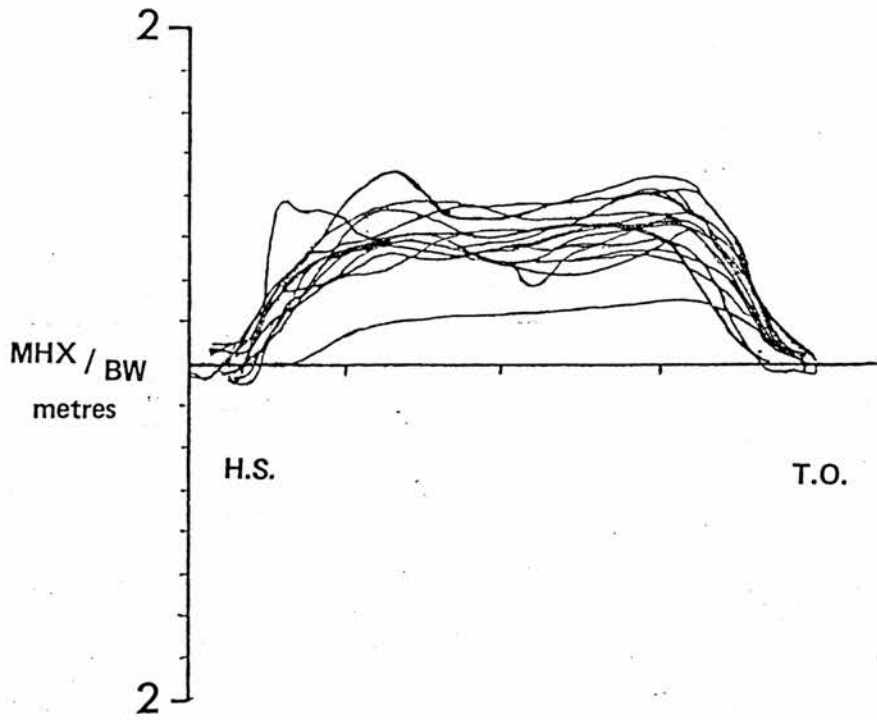


Fig. 4.42 Resultant Moments about Replaced Hip Joints.

HEALTHY LIMB
(Post - op)

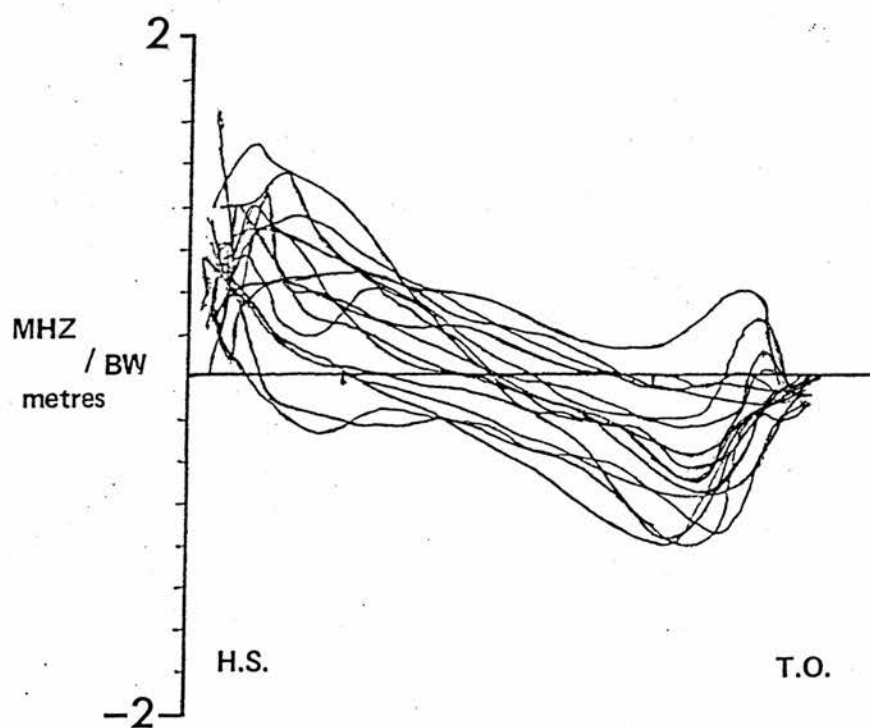
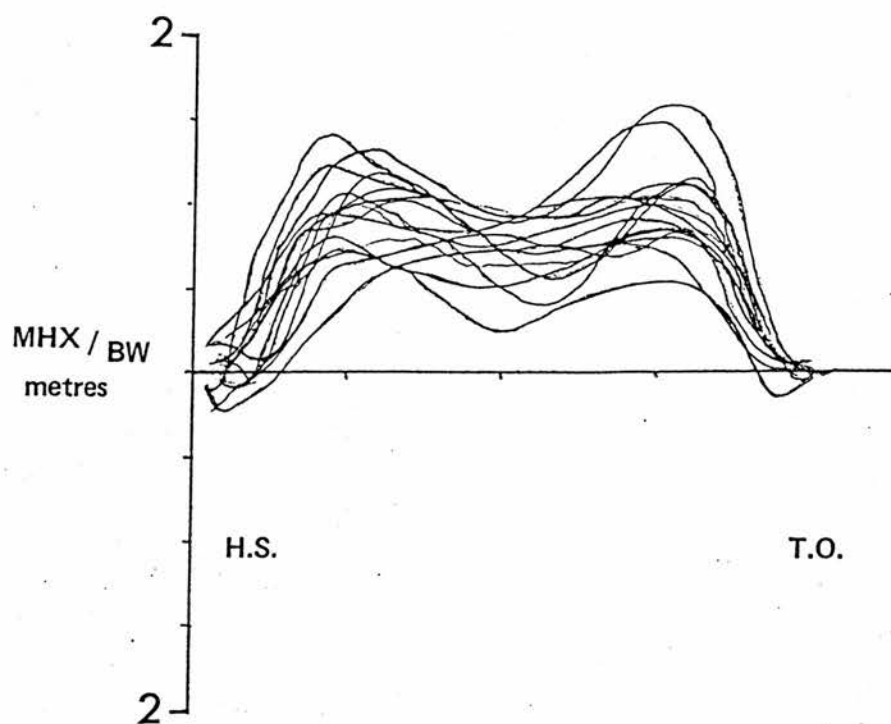


Fig. 4.43 Resultant Moments about Healthy Hip Joints.

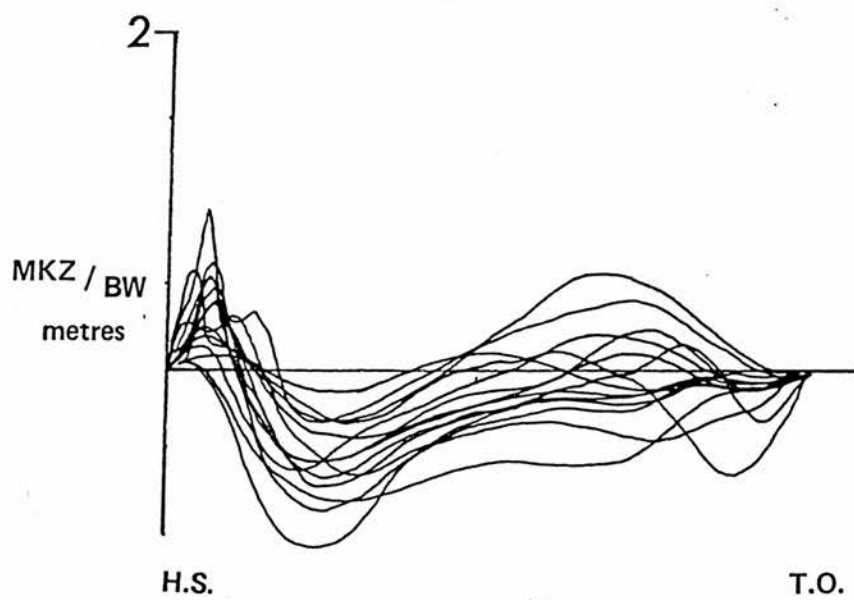
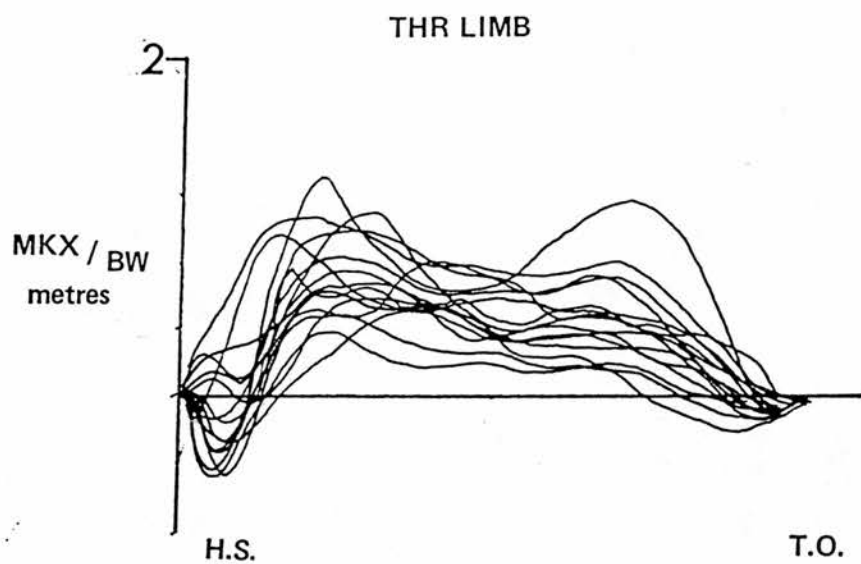


Fig. 4.44 Resultant Moments about
Ipsilateral Knee Joints.

HEALTHY LIMB
(Post - op)

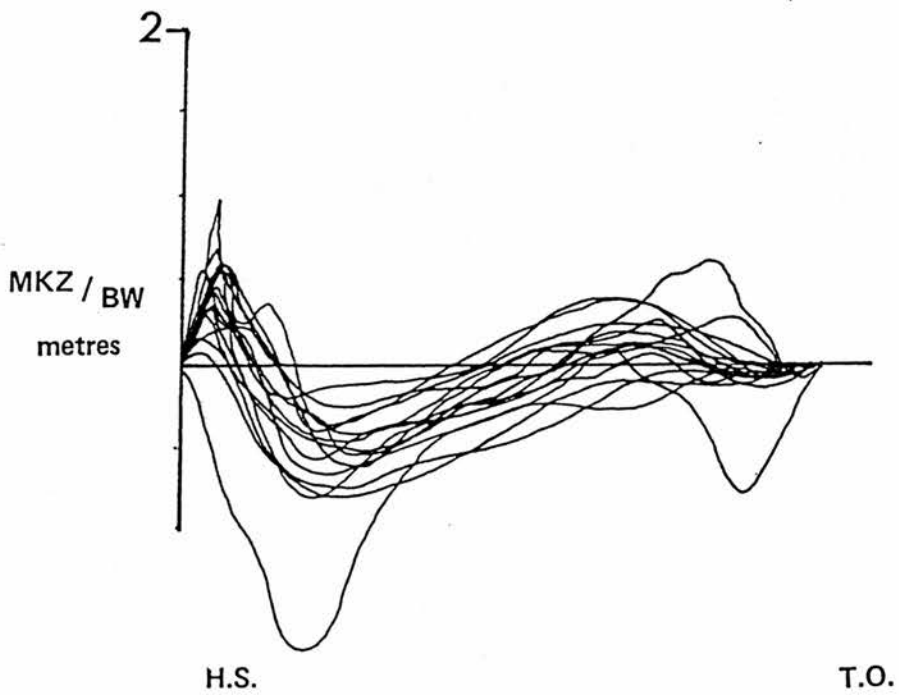
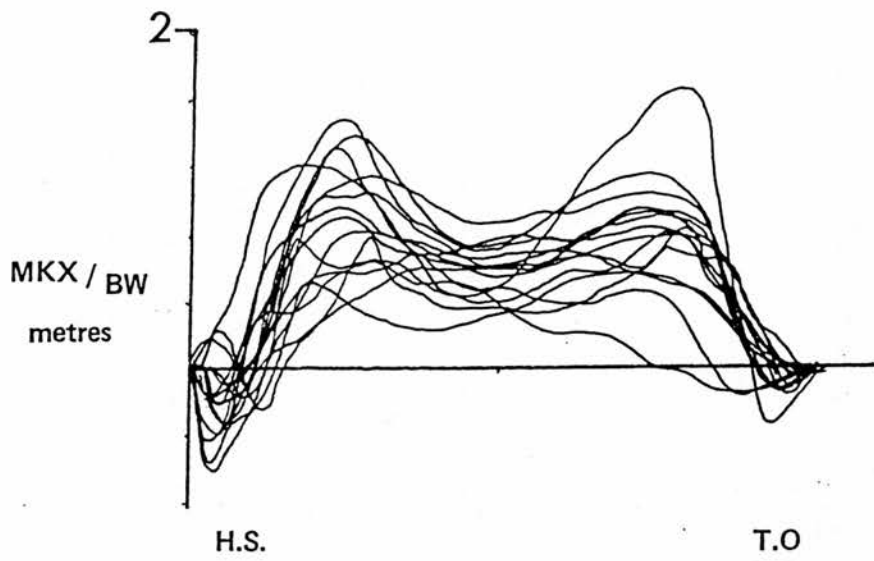


Fig. 4.45 Resultant Moments about
Contralateral Knee Joints.

ipsilateral operated limb knee. Here, the pattern has become very similar to that seen at the healthy knee although the initial and terminal positive deflections are usually of lower magnitude indicating less need for hamstrings and gastrocnemius action and reflecting the recovery in the movement pattern at this knee (see Figs. 4.10 and 4.38).

e. Joint Loads

Knee Joint (Figs. 4.46 and 4.47)

In the healthy limbs the three peak pattern is seen, usually with a larger terminal peak. In the ipsilateral knee joints there is a return towards the "normal" pattern but this is less marked than at the contralateral knees.

The magnitude of the peak loads across the knees differs, that across the contralateral knees being 3.16 ± 1.66 xBW and across the ipsilateral knees 2.61 ± 1.64 xBW ($p = 0.05$ paired t-test).

When the compartmental loading patterns are considered, a considerable variation is seen (Table 4.2). At the contralateral knees, 23.31% and 6.54% of the total knee joint load is transmitted by the lateral compartment compared with $29.88 \pm 7.80\%$ at the ipsilateral knees. The contralateral knees therefore show little overall change in load distribution from the pre-operative state whilst the ipsilateral knees carry more load through the lateral compartment.

Hip Joints

The hip joint loads for the group are shown in Figures 4.48 and 4.49.

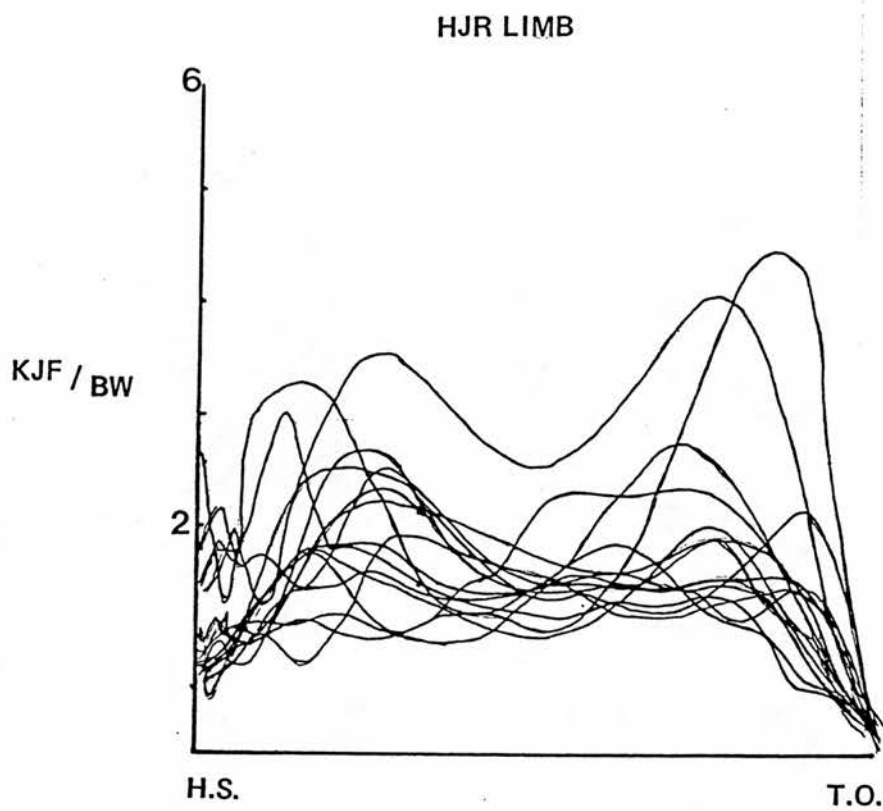


Fig. 4.46 Knee Joint Loads - HJR Limbs.

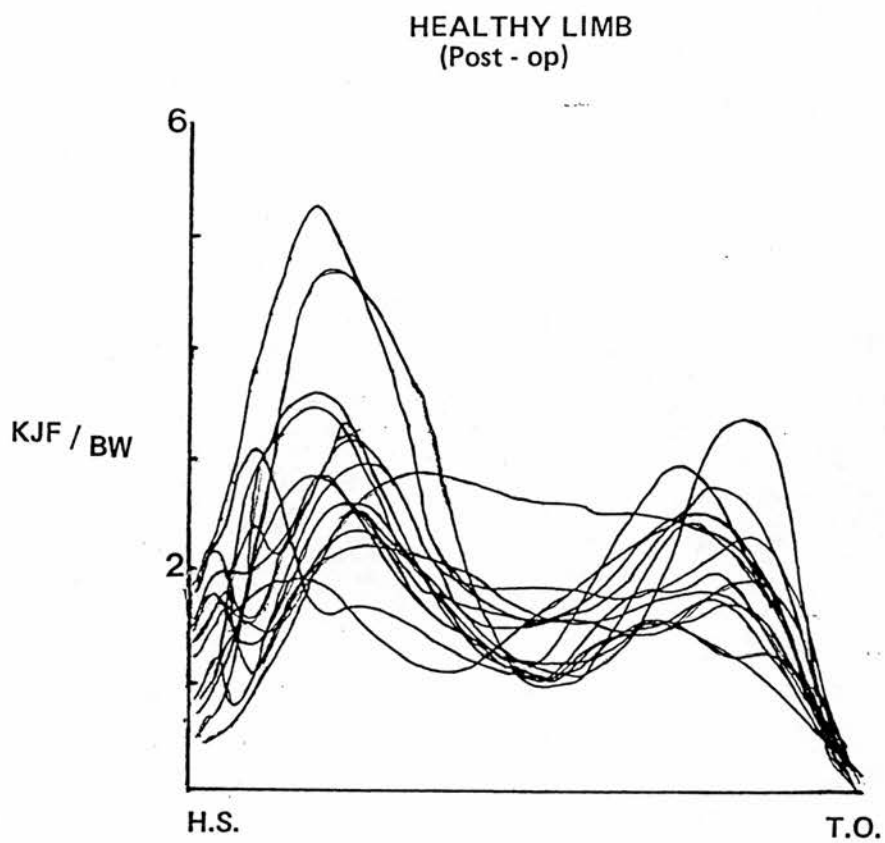


Fig. 4.47 Knee Joint Loads - Healthy Limbs.

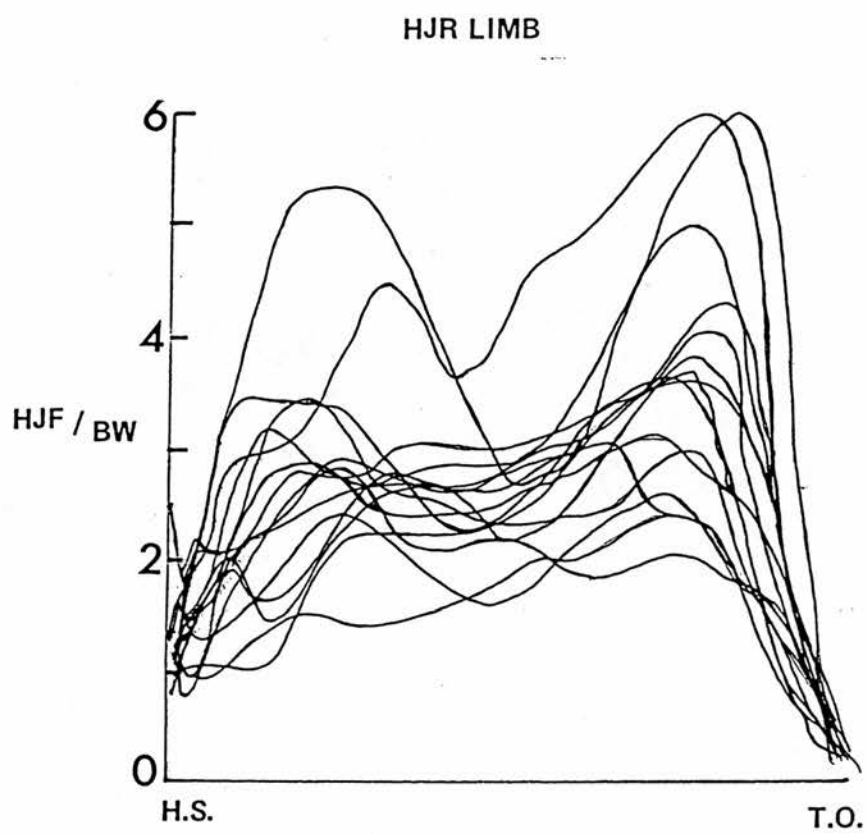


Fig. 4.48 Hip Joint Loads - HJR Limbs.

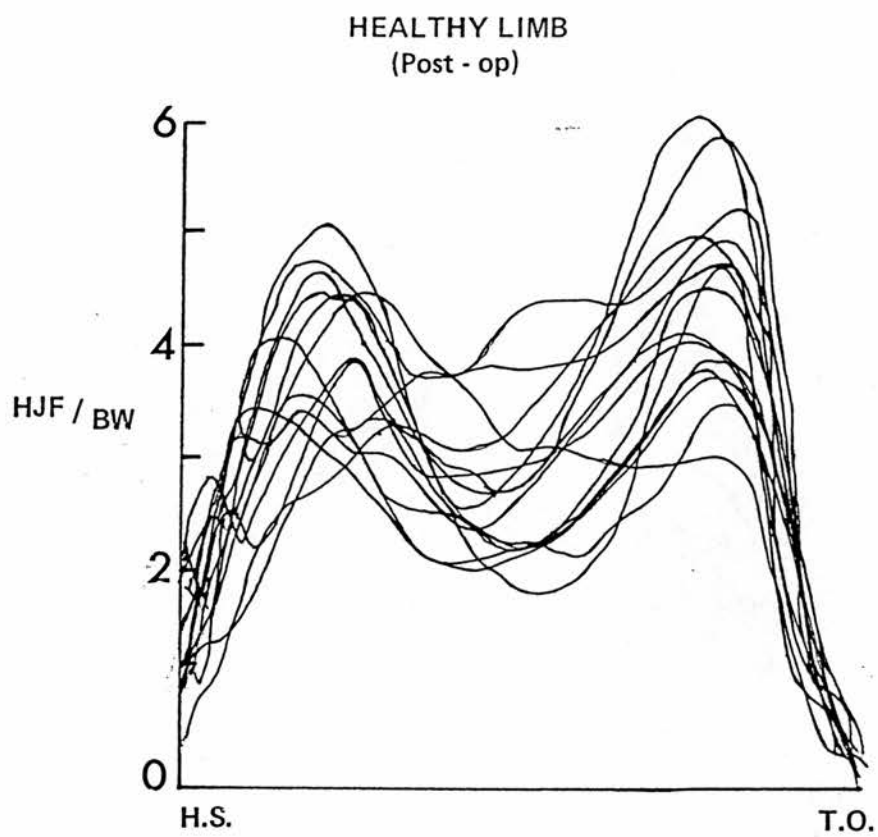


Fig. 4.49 Hip Joint Loads - Healthy Limbs.

At the healthy hips there is a similar degree of scatter to that seen in a group of normal subjects and the magnitude is similar at $4.73 \pm 1.30 \times \text{BW}$. Most patients show a two peak curve but three patterns can be recognised: a post heel-strike peak of greater magnitude, a pre toe-off peak of greater magnitude, or two peaks of similar magnitude. The first pattern results from greater force of contraction in the hip extensors (gluteus maximus and the hamstrings) and abductors than in the flexors and abductors. The second pattern is the reverse of this and the third is obviously produced by more equal force of contraction in the muscle groups. These patterns are all seen in normal subjects (see Fig. 1.8) and probably here reflect the individual's mode of gait which result from posture and the use of the upper as well as the lower limbs. This produces those features which make a person recognisable by his gait. In the patients studied features such as pain and limitation of movement (see Chapter 5) are probably more significant in the production of these patterns. No consistency of pattern from pre- to post- operatively was noted.

At the replaced hip joints a wide scatter of magnitudes of joint load are seen with a mean of $3.86 \pm 1.76 \times \text{BW}$ ($p = 0.001$ paired t-test). Some curves show two peaked forces but the majority exhibit irregular plateaux with a rise to a maximum prior to toe-off, a feature indicative of reduced abductor and/or extensor muscle activity in early stance phase.

A difference between the groups treated with different arthroplasties should be noted here. For those with C.A.D. Müller replacements the mean force at the replaced hip was 4.27 ± 1.46 xBW and at the healthy hip 4.79 ± 1.22 xBW ($p = 0.05$ paired t-test). When the Charnley arthroplasty was used the mean force at the replaced hip was 3.52 ± 1.84 xBW and at the healthy hip 4.69 ± 1.48 xBW ($p = 0.001$ paired t-test).

f. Phasing of Hip Joint Force Maxima

Pre-operatively the phasing of the heel strike and toe-off maxima in the hip joint force are as shown in Table 4.2

Post-operatively the phasing changes especially for the heel strike peak at the replaced hips as shown in Table 4.2

If it is assumed that post-operatively stance phase occupies 60 per cent of the total gait cycle for each limb (Paul, 1967) then the heel strike peaks occur at 12.6 and 13.8 per cent for the healthy and operated limb respectively, and 46.2 and 45.6 per cent for the toe-off peak. This compares well to Paul's figures of 7 per cent and 47 per cent for the normal subject (Paul, 1967).

TABLE 4.2a

BIOMECHANICAL RESULTS - HIP DISEASE GROUP

A = Affected Limb
H = Healthy Limb

Patient	Stance Phase Durn. (secs)	Av. Forward Velocity m.s	MHX (max)	MHZ (max) metres	MKX (max)	MKZ (max)	Muscle Forces			HJF (max) xBW	Phase HS TO %Stance	KJF (max) xBW	% Lat. Compartment Load		
1	A 0.68	1.65	0.45	0.60	0.20	0.45	1.1	0.6	0.3	0.8	2.3	30	60	1.7	25
	H 0.76		1.15	0.85	0.15	0.80	2.7	1.4	1.2	1.8	4.8	11	75	2.9	44
2	A 0.56	0.57	0.45	0.23	0.45	0.10	1.1	0.3	0.3	0.2	2.2	-	-	1.7	10
	H 0.78		0.90	0.65	0.55	0.40	2.4	1.3	0.0	0.9	3.8	-	-	2.1	25
3	A 0.70	0.87	0.70	0.75	0.55	0.40	2.0	1.6	1.0	0.9	4.0	20	80	2.2	13
	H 0.84		0.75	0.50	0.25	0.40	1.8	0.7	2.0	1.0	4.1	16	75	3.4	30
4	A 0.82	1.11	1.00	0.30	0.35	0.80	0.6	0.3	1.9	1.7	3.3	34	78	2.7	17
	H 0.84		1.25	0.70	0.50	0.80	2.3	0.8	1.5	1.9	4.6	15	75	2.8	14
5	A 0.90	0.60	0.00	0.15	0.15	0.40	0.1	0.1	1.8	0.3	2.4	40	75	2.5	66
	H 0.94		1.30	0.60	0.60	0.40	3.0	1.2	0.6	0.7	4.7	20	72	1.0	4
6	A 0.72	0.61	0.32	0.30	0.15	0.50	0.1	0.3	0.7	1.7	1.5	20	58	2.7	36
	H 0.90		1.15	0.35	0.38	0.90	1.4	1.2	0.8	2.1	3.0	12	70	3.1	28
7	A 0.78	0.97	0.95	0.50	0.35	0.38	1.4	0.2	1.1	0.8	3.4	24	75	1.8	36
	H 0.70		0.75	0.75	0.66	0.65	1.8	2.3	0.5	0.7	4.6	20	78	3.2	28
8	A 0.82	0.77	1.00	0.60	0.68	0.80	2.0	0.6	0.9	2.0	3.4	30	76	3.0	6
	H 0.94		1.10	0.85	0.55	0.25	2.0	2.3	0.5	0.6	4.6	18	75	2.2	13

TABLE 4.2a (continued)

BIOMECHANICAL RESULTS - HIP DISEASE GROUP

A = Affected Limb

H = Healthy Limb

Patient	Stance Phase Durn. (secs)	Av. Forward Velocity m.s	MHX (max)	MHZ (max) metres	MKX (max)	MKZ (max)	Muscle Forces xBW			HJF (max) xBW	Phase HS TO %Stance	KJF (max) xBW	% Lat. Compartment Load	
9	A 0.76	0.91	0.72	0.35	0.15	0.28	1.2	0.6	0.3	0.3	2.8	32 80	2.1	50
	H 0.80		0.90	1.20	0.35	0.75	2.0	0.7	3.2	2.0	5.9	30 75	3.0	24
10	A 0.70	0.90	0.60	0.78	0.38	0.45	2.1	0.2	2.0	1.4	4.0	30 70	2.1	24
	H 0.76		1.22	1.15	0.41	0.70	3.3	2.3	1.1	1.5	5.7	12 75	2.3	30
11	A 0.66	0.75	0.63	0.62	0.20	0.72	1.9	0.7	0.4	0.1	3.3	35 80	2.3	43
	H 0.90		0.82	1.10	0.22	0.65	2.1	0.5	2.8	1.6	5.3	15 72	2.7	29
12	A 0.74	0.79	0.90	0.38	0.40	0.20	1.9	0.9	0.4	0.5	3.2	30 62	1.6	26
	H 0.76		0.78	0.95	0.38	0.28	1.7	0.4	2.1	0.9	4.2	25 70	2.0	15
13	A 1.04	0.59	0.45	0.45	0.20	0.71	1.1	0.9	1.0	1.5	2.7	25 75	2.4	55
	H 1.08		1.30	1.10	0.55	1.01	3.0	0.4	2.3	2.6	5.8	30 75	3.2	20
14	A 0.66	1.03	0.70	0.45	0.60	0.28	1.7	0.8	0.2	0.7	2.8	25 74	2.1	20
	H 0.74		1.25	1.20	0.42	1.45	3.2	2.9	0.8	2.8	5.7	17 77	3.7	17
15	A 0.76	0.84	0.80	0.50	0.55	0.20	1.5	0.4	1.0	1.0	3.2	35 70	2.3	13
	H 0.82		1.90	0.55	0.53	0.60	2.7	1.6	2.1	1.5	5.2	17 76	2.7	12
16	A 0.64	1.12	0.50	0.55	0.38	0.25	2.7	0.4	2.0	0.8	5.1	22 70	1.9	29
	H 0.68		1.50	0.65	0.92	1.60	4.8	1.1	2.6	3.9	6.0	12 78	5.2	12

TABLE 4.2b
BIOMECHANICAL RESULTS - HIP REPLACEMENT GROUP
R = Replaced joint limb H = Healthy Limb

Patient	Stance Phase Durn. (secs)	Av. Forward Velocity m.s	MHX (max)	MHZ (max) metres	MKX (max)	MKZ (max)	Muscle Forces xBW				HJF (max) xBW	Phase HS TO %Stance	KJF (max) xBW	% Lat. Compartment Load	
1	R 0.74	1.69	1.00	0.55	0.60	0.70	2.6	1.4	1.3	1.5	4.7	20	78	2.8	9
	H 0.72		0.80	1.00	0.25	1.43	1.9	1.3	2.1	3.1	4.5	20	80	4.0	40
2	R 0.74	0.88	1.00	0.75	0.60	0.55	1.5	0.7	1.7	1.3	3.7	20	75	2.3	13
	H 0.78		0.88	0.70	0.85	0.76	2.3	2.0	1.1	1.7	4.5	30	81	3.2	6
3	R 0.68	1.05	0.65	0.37	0.20	0.30	1.8	0.5	1.1	0.6	3.7	18	75	2.1	66
	H 0.72		1.05	0.85	0.40	1.10	2.5	1.2	2.7	2.5	5.8	18	78	3.5	22
4	R 0.78	1.28	1.10	0.80	0.38	1.85	0.3	0.8	4.8	4.3	5.7	35	76	5.5	40
	H 0.80		1.55	1.20	0.50	0.28	3.1	1.7	2.4	0.7	6.0	22	75	3.2	26
5	R 0.56	1.27	0.93	1.40	0.50	1.00	2.0	0.6	1.6	1.2	4.1	25	82	2.4	27
	H 0.58		1.10	0.65	0.97	0.85	3.0	1.5	1.7	2.5	4.7	24	80	4.7	3
6	R 0.70	1.16	1.00	0.75	0.60	0.40	1.1	0.8	0.8	1.0	2.7	25	75	2.5	20
	H 0.74		1.60	1.15	1.08	0.55	2.3	1.2	1.0	1.2	4.3	15	77	3.4	9
7	R 0.66	0.95	0.72	0.60	0.25	0.30	1.3	0.8	0.6	0.4	2.9	25	78	2.1	43
	H 0.74		1.25	0.75	0.58	0.80	2.5	1.5	1.2	1.7	4.3	15	80	3.0	17
8	R 0.64	0.95	0.90	0.45	0.40	0.45	1.8	0.6	1.2	1.1	3.6	33	75	2.0	14
	H 0.72		1.15	0.80	0.70	0.70	2.1	0.6	1.9	1.6	4.2	30	72	2.9	23

TABLE 4.2b (continued) BIOMECHANICAL RESULTS - HIP REPLACEMENT GROUP
R = Replaced joint limb H = Healthy Limb

Patient	Stance Phase Durn. (secs)	Av. Forward Velocity m.s	MHX (max)	MHZ (max)	MKX (max)	MKZ (max)	Muscle Forces			HJF (max) xBW	Phase HS TO %Stance	KJF (max) xBW	% Lat. Compartment Load		
9	R 0.68	1.21	0.80	1.50	0.30	0.75	1.0	0.6	3.7	2.3	5.0	26	79	3.2	32
	H 0.66		1.05	1.10	0.34	0.80	2.9	1.1	3.1	2.1	6.0	24	78	3.4	30
10	R 0.66	1.00	0.78	0.50	0.50	0.35	1.9	0.6	0.3	0.6	3.1	25	67	1.7	24
	H 0.66		1.15	0.75	0.38	0.55	3.3	1.4	0.6	0.3	5.6	18	80	2.8	35
12	R 0.70	0.95	1.30	0.60	0.10	0.50	3.6	0.7	1.8	1.2	5.0	30	75	2.3	50
	H 0.70		1.10	1.15	0.38	0.55	2.3	0.6	2.6	1.3	5.0	30	74	2.4	20
13	R 0.70	1.10	0.90	0.60	0.38	0.60	1.9	1.3	0.6	0.8	3.8	16	76	2.9	29
	H 0.70		1.20	1.30	0.40	0.65	3.3	0.6	1.5	1.4	4.8	30	80	2.4	22
14	R 0.62	1.25	1.15	0.90	0.55	0.65	2.8	1.0	0.6	0.6	4.0	17	74	2.9	26
	H 0.66		1.30	0.78	0.61	0.80	2.9	1.5	1.6	1.8	4.7	20	78	3.2	15
15	R 0.56	1.40	0.85	1.20	0.34	0.65	2.2	0.3	2.6	1.6	5.4	25	76	3.7	18
	H 0.56		1.10	0.70	0.72	1.45	3.8	1.3	2.0	3.4	6.0	20	80	4.7	15
16	R 0.62	1.25	0.80	0.30	0.30	0.51	1.9	0.6	1.1	0.9	3.8	15	74	3.2	39
	H 0.68		0.65	1.40	0.90	1.7	2.8	0.5	3.4	4.1	5.8	20	80	5.2	33
25	R 0.76	1.11	0.40	0.70	0.28	0.45	1.3	1.3	1.4	0.9	3.1	24	72	2.7	35
	H 0.80		1.10	0.60	0.25	0.75	2.6	1.2	1.2	1.8	4.5	24	76	2.8	50

4.IV GIRDLESTONE EXCISION ARTHROPLASTY GROUP

This group comprised nine patients one of whom later underwent conversion to a total hip replacement. The clinical details of these patients appear in Table 4.3

1. Results

The results for the entire group are presented in Table 4.4 at the end of this section and because of the heterogenicity of the group the mean results for the entire group will be considered in relation to those of the pre- and post-operative hip replacement groups under the following headings:-

- a. Time-Distance Parameters
- b. Limb Movements
- c. Ground-Foot Reaction Forces
- d. Resultant Moments at the Hip and Knee
- e. Muscle and Joint Forces

a. Time-Distance Parameters

Stance phase duration was 0.86 ± 0.24 sec. on the non-affected side and 0.78 ± 0.20 sec. on the Girdlestone (GS) side. This represents a 10 per cent reduction in the duration of stance on the GS limbs compared to the unaffected limbs.

The walking velocity of this group of patients was not measured. The duration of stance phase suggests that the velocity was probably similar to that of the hip disease group.

b. Limb Movements

The movements at both the hip and knee joints are shown in Figures 4.50 and 4.51. The patterns of movement are assymmetrical and resemble those seen in the degenerative hip disease group.

Hip Joint

At the hip the range of movement used during walking on the GS side is approximately two-thirds of that on the unaffected side principally as a result of reduced flexion following heel strike. The mean range at the GS hips is from 12.5° flexion to 14° extension; a smaller range than that seen in the pre-operative group of hip replacement patients, and much smaller than that used by the normal subjects. Again the change from flexion to extension occurs at 40 per cent stance.

At the unaffected hip the range of movement was from 26° of flexion to 10° extension, the changeover occurring at approximately 60 per cent of stance phase.

Knee Joint

The differences in the pattern of movement at the knee joints is extremely marked. The unaffected limb retains the triphasic pattern seen in the normal subjects but remains very much more flexed throughout stance phase with a post heel strike peak of 28° and a mid-stance minimum of 17° .

The knee of the GS limb shows a pattern very similar to that seen in the ipsilateral knee joint of the patients with unilateral osteoarthritis of the hip, with no post heel strike flexion and almost full extension until beyond 60 per cent of stance phase.

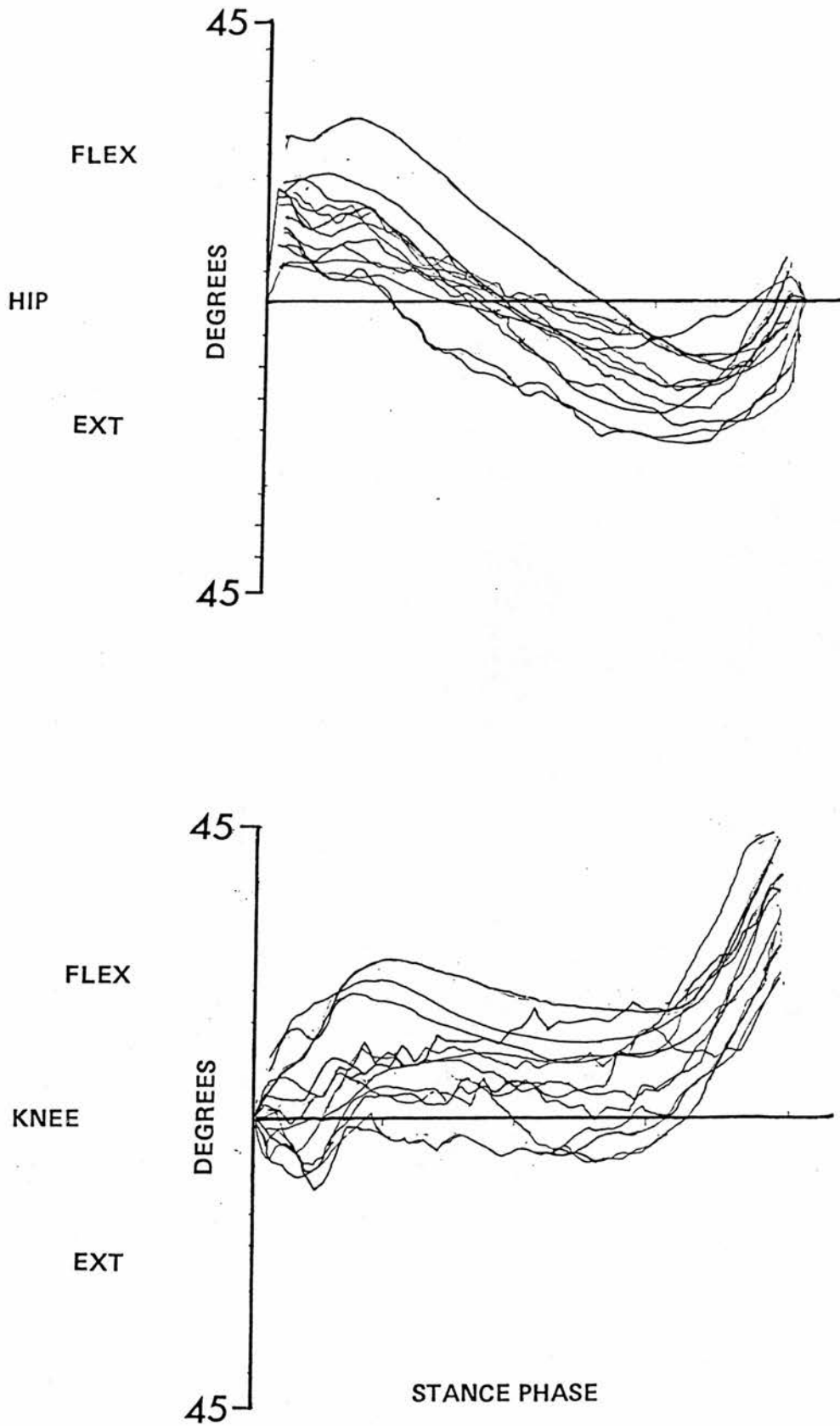


Fig. 4.50 Range of Movement at Hip
and Knee Joints - GS Limbs.

HEALTHY LIMB

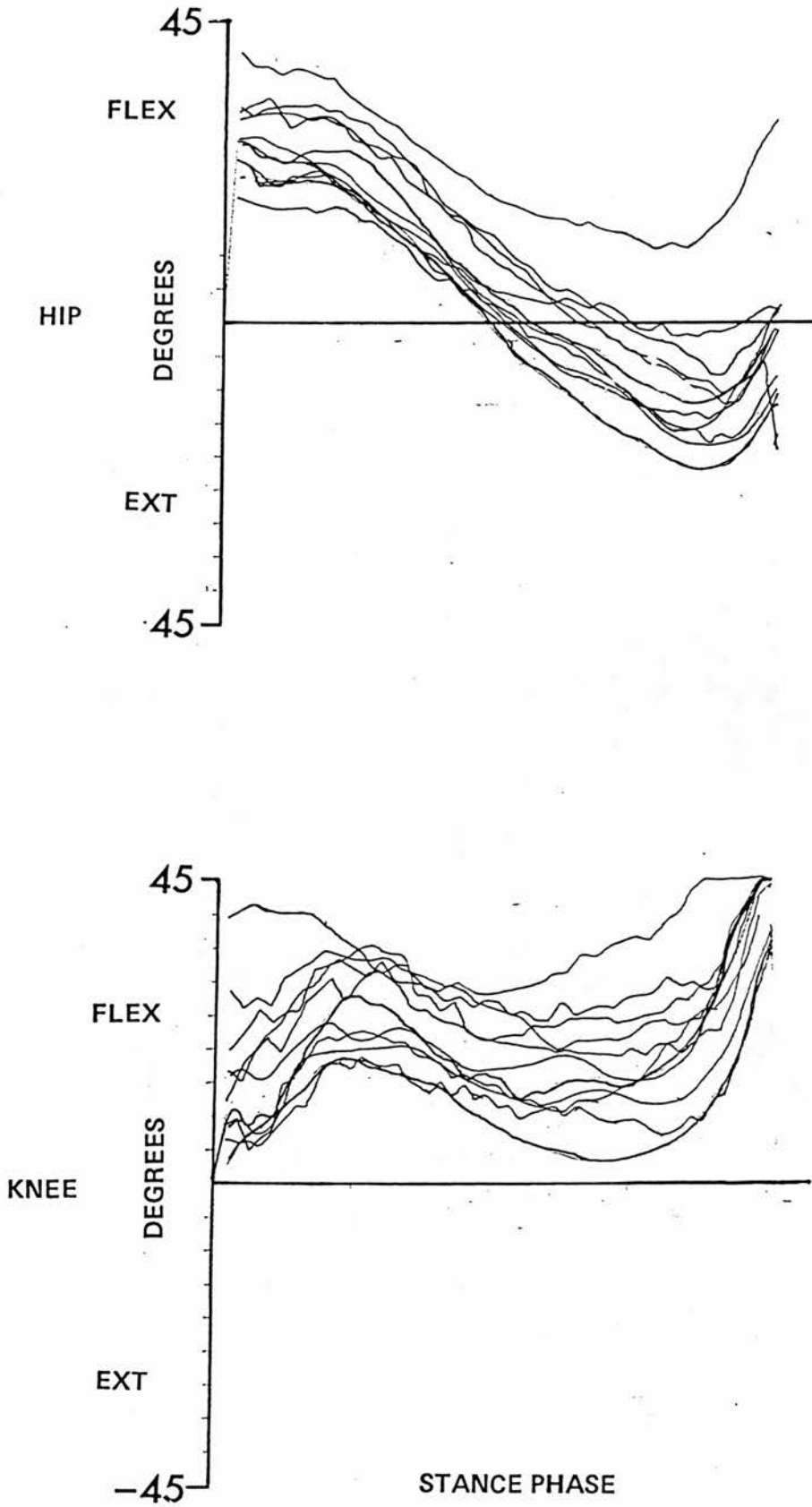


Fig. 4.51 Range of Movement at Hip and Knee Joints
- Healthy Limbs.

c. Ground-Foot Reaction Forces

In the fore and aft direction FXG shows the typical pattern seen in Figures 4.52 and 4.53 being of greater magnitude on the unaffected limb especially towards toe-off. However, the magnitude of the forces acting on both limbs is less than that seen in the normal subjects. The forces on the unaffected limb compare well with those on the healthy limb of the hip disease group but the forces on the GS limb are very much smaller than those on the affected side in that group. Only the unaffected limb of the Girdlestone group of patients shows the sharp initial peak in FXG.

The vertical component of the ground reaction force (FYG) is similar for both limbs with less of the normal bimodal pattern. However, the GS limb shows earlier unloading than the unaffected limb.

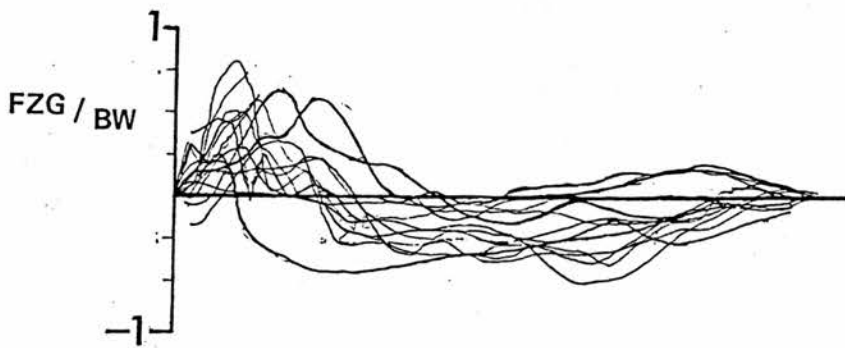
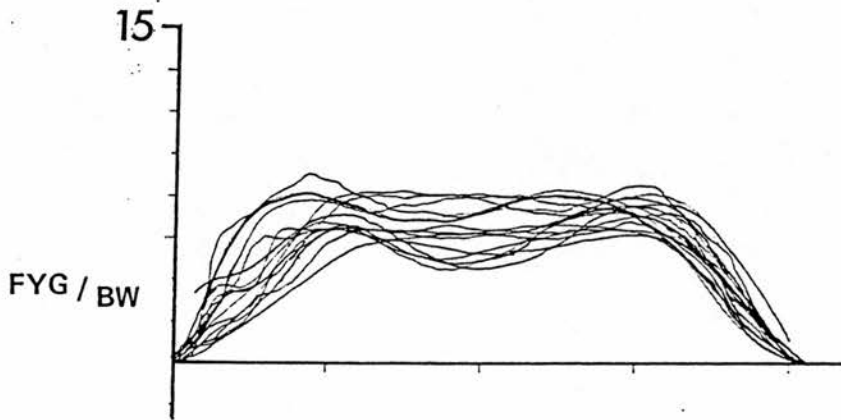
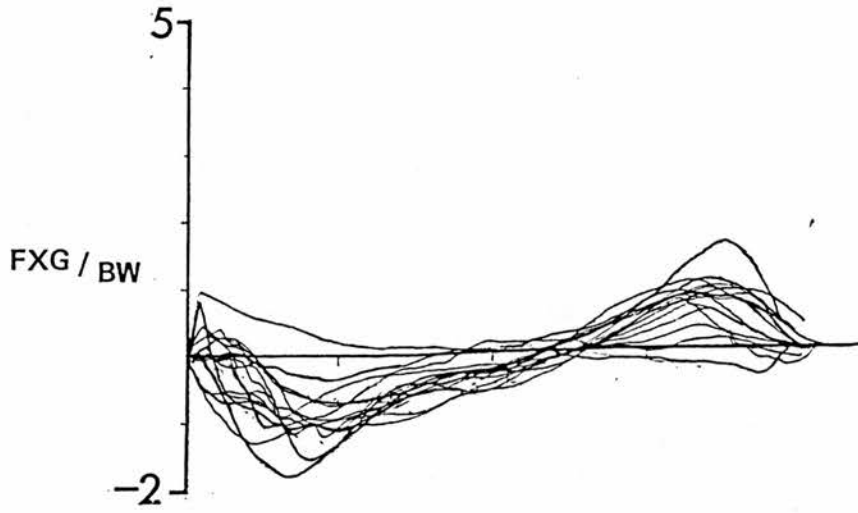
The mediolateral component of the ground force (FZG), shows marked differences between the limbs. The pattern seen on the unaffected side closely resembles that seen in the hip disease group and is similar to that seen in the normal subjects although with a more marked terminal medial force. The pattern of the force acting on the GS limb differs in that the initial laterally directed force persists for 30 per cent of stance phase.

d. Resultant Moments at the Hip and Knee

Hip Joint

The moments acting about both hip joints are shown in Figures 4.54 and 4.55. There is a reduction in the

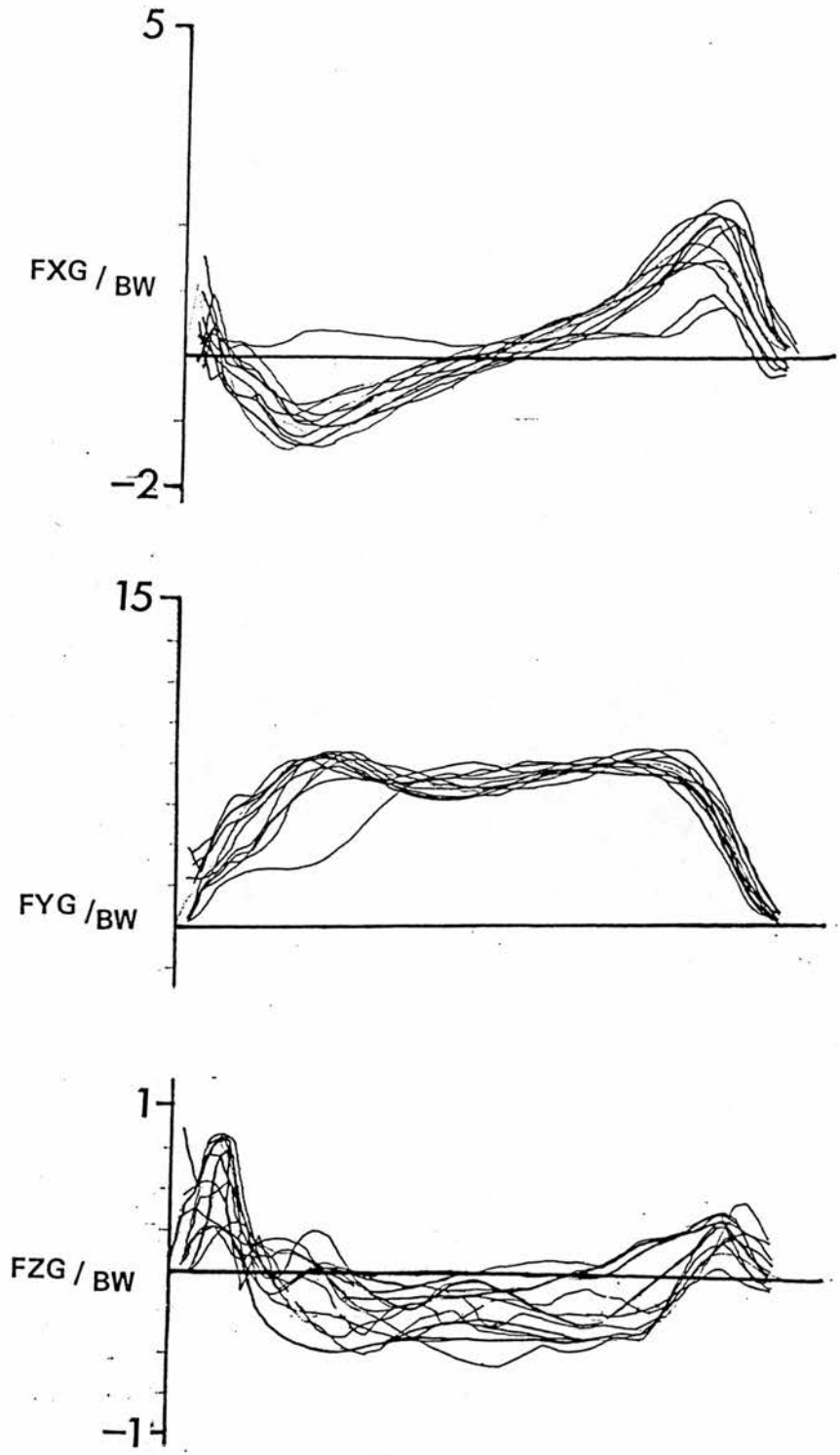
GS LIMB



STANCE PHASE

Fig. 4.52 Ground Reaction Forces - GS Limbs.

HEALTHY LIMB



STANCE PHASE

Fig. 4.53 Ground Reaction Forces - Healthy Limbs.

GS LIMB

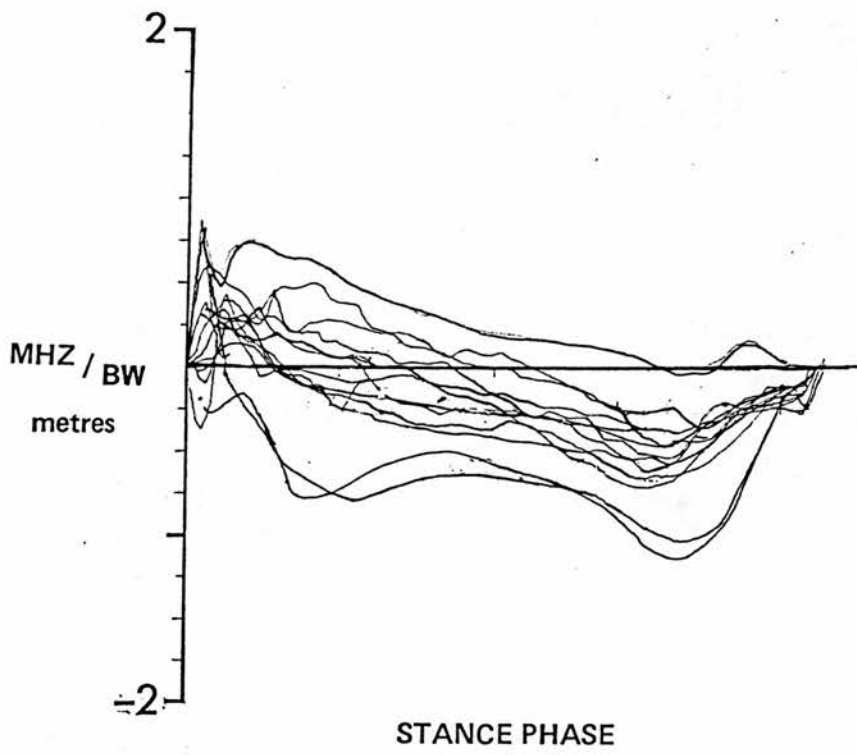
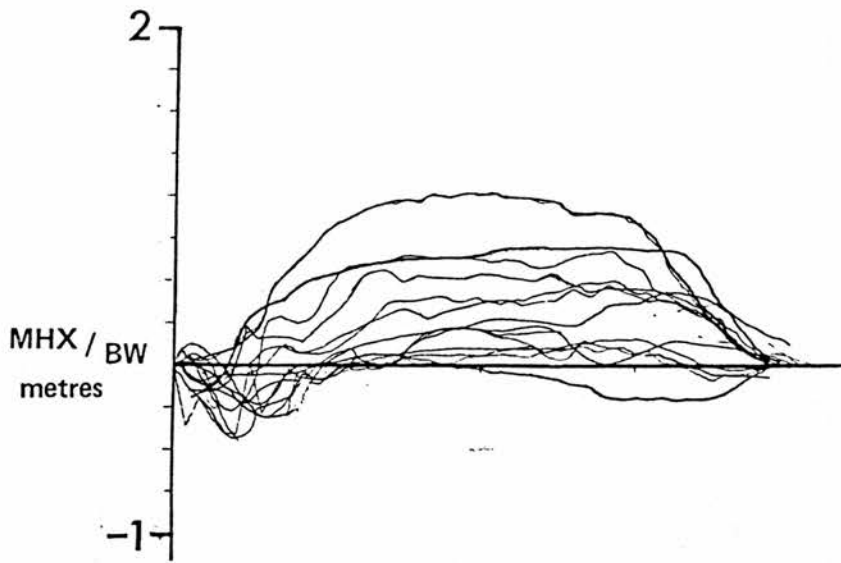


Fig. 4.54 Resultant Moments about the GS Hip Joints.

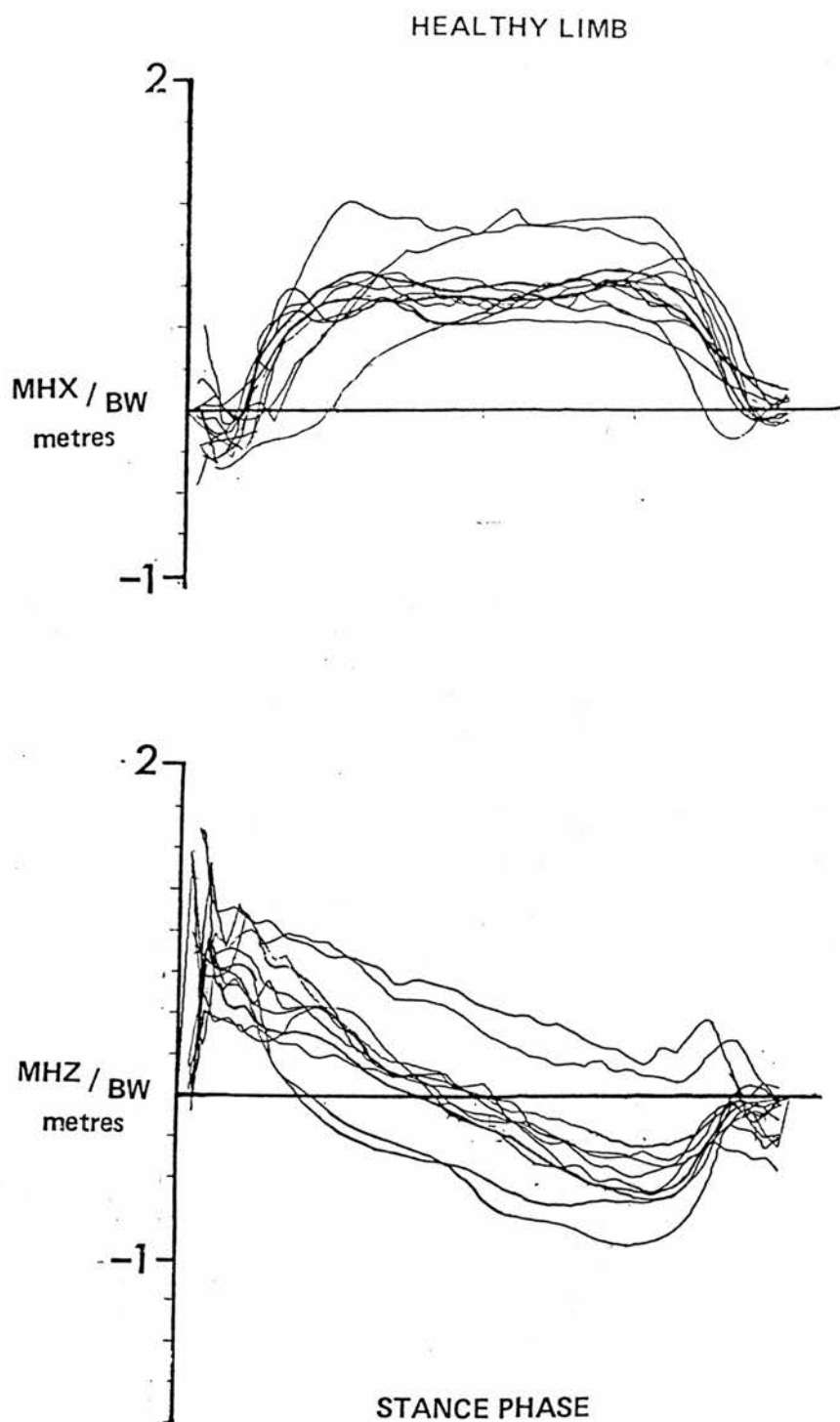


Fig. 4.55 Resultant Moments about
the Healthy Hip Joints.

maximum value of all moments acting about the GS limb.

The abduction-adduction moment (MHX) lacks the typical two trough and peak pattern in both limbs and both limbs also show a marked initial negative deflection indicative of adductor muscle activity.

The flexion-extension moment (MHZ) becomes negative earlier in the GS limb corresponding to the earlier move into extension and is more negative (suggesting activity in the extensor muscles) until late stance. The moment about the healthy hip is very similar to that about the hips of normal subjects.

Knee Joint

Only moments about the X axis at the knees (MKX) are available for this group (Figs. 4.56 and 4.57). As mentioned earlier these moments are opposed by force actions in the collateral ligaments and by joint contact forces and the sign and magnitude of MKX will be an indication of the compartmental distribution of load at the knee joint.

An initial negative deflection is seen after heel strike in both limbs reflecting the laterally directed FZG at this phase. Thereafter all of the contralateral knees show a positive deflection of varying magnitude indicating medial joint compartment loading. However, several of the ipsilateral knees show little or no positive deflection at all indicating load carried almost totally within the lateral compartment of that knee joint. Also in the ipsilateral knees the magnitude of positive deflections, when present, is smaller in most

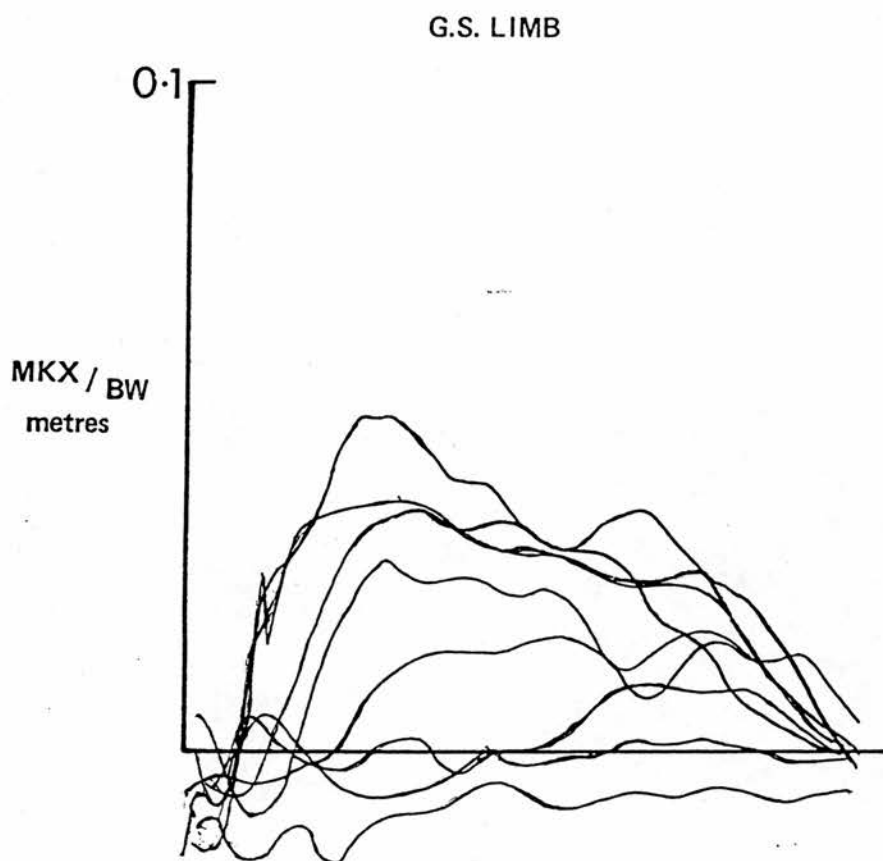


Fig. 4.56 Resultant Moments about the Axis of
Progression at the Ipsilateral Knee Joints.

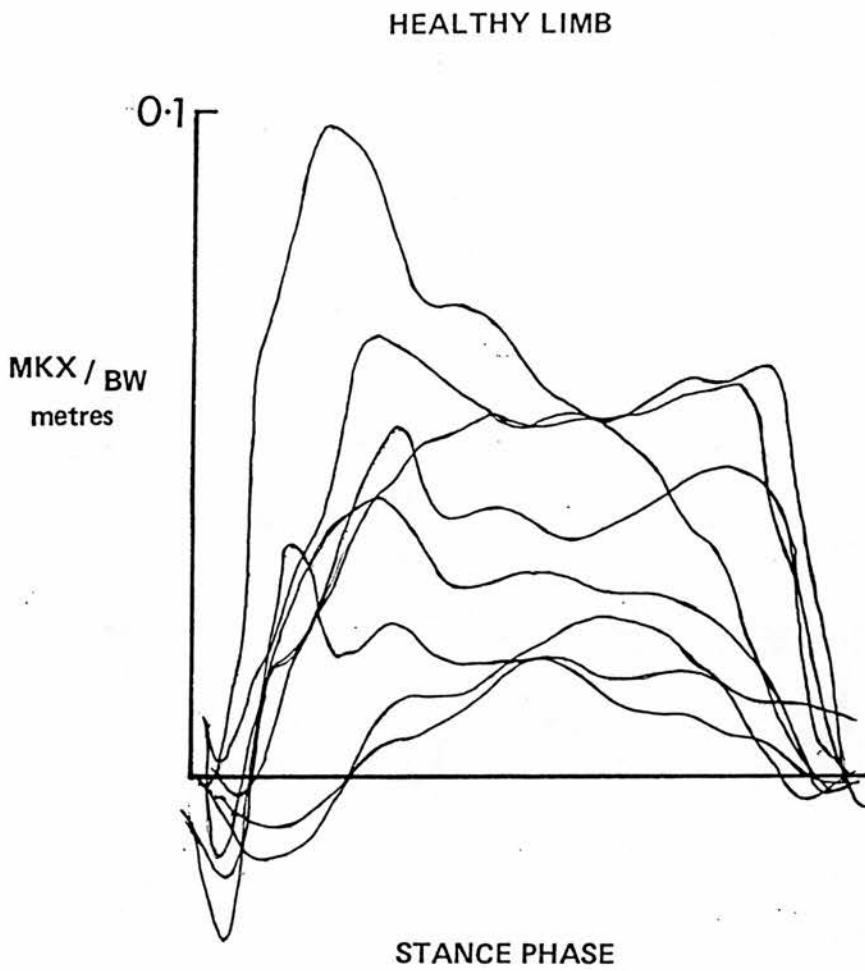


Fig. 4.57 Resultant Moments about the Axis of Progression at the Healthy Contralateral Knee Joints.

cases, than that seen in the contralateral knees.

e. Muscle and Joint Forces

The magnitudes of the muscle and joint forces varied greatly within the group especially in the affected limb and, in part, this reflected the use of a walking stick by some patients. Brand and Crowninshield (1980) has shown that a reduction of up to 60 per cent of hip joint load can be achieved by the use of a stick.

Hip Joint

At the hip joints greater loads were usually seen in the unaffected limbs (Fig. 4.58). The same two patterns of loading were seen with varying influences of abductor and flexor/extensor muscles (Table 4.4), but there was a definite tendency for the affected hips to show a peak towards toe-off suggesting inefficacy of abductor musculature in early stance phase.

Knee Joint

The differences in the loads across the knees were much less marked and were not as great as those seen after total hip arthroplasty (Fig. 4.59).

The most notable feature was the virtual absence of load attributable to quadriceps contraction in the knees of the affected limbs. This was presumably related to their movement pattern and emphasises the part played by the gastrocnemius muscle in the generation of knee joint force in these patients.

The compartmental loading pattern, as suggested by MKX, showed differences between ipsilateral and contralateral knees. A very much greater proportion of the load is

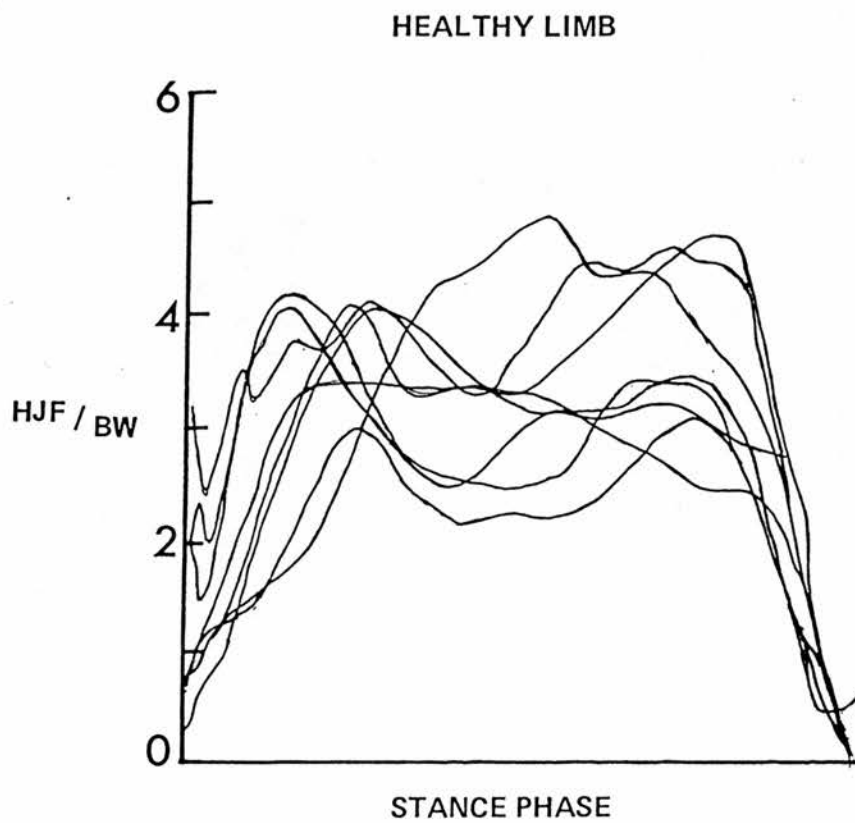
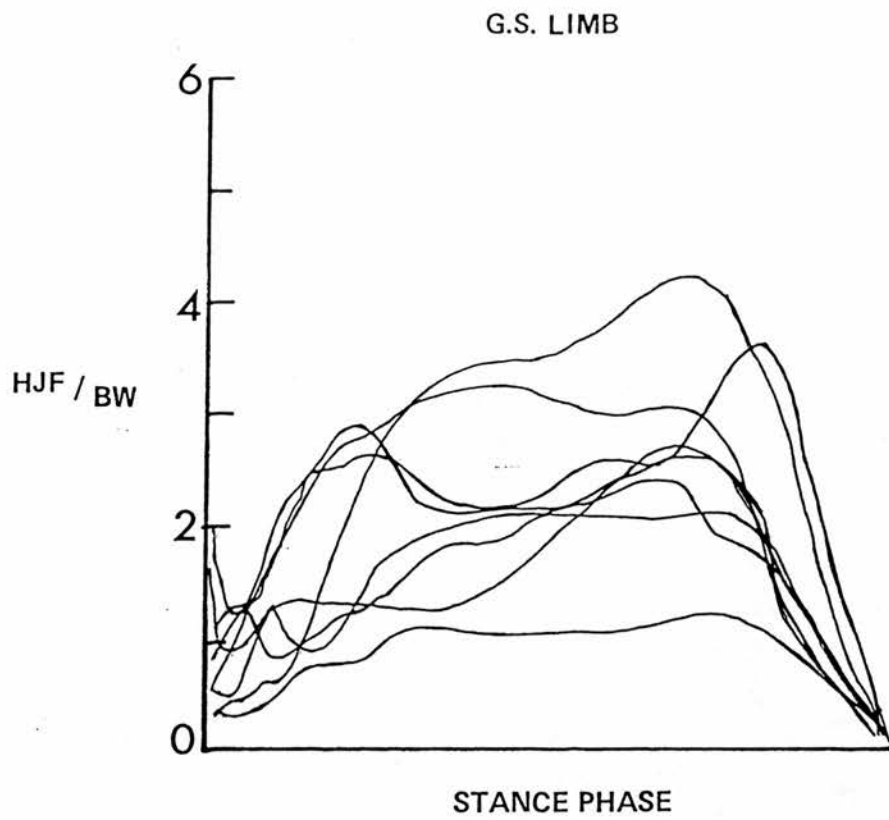


Fig. 4.58 Hip Joint Loads.

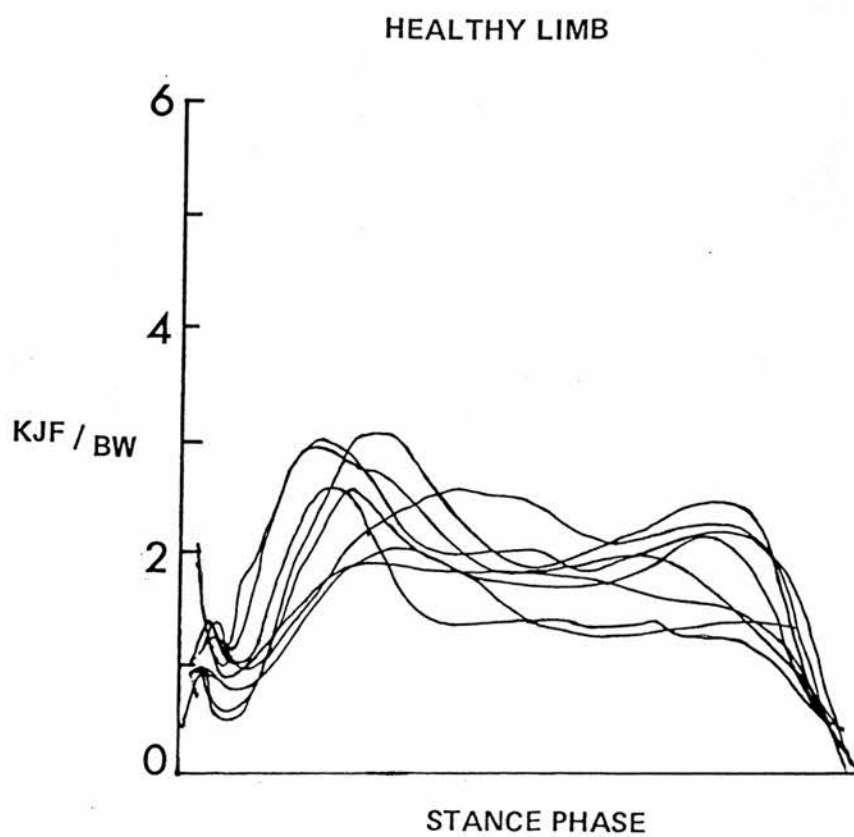
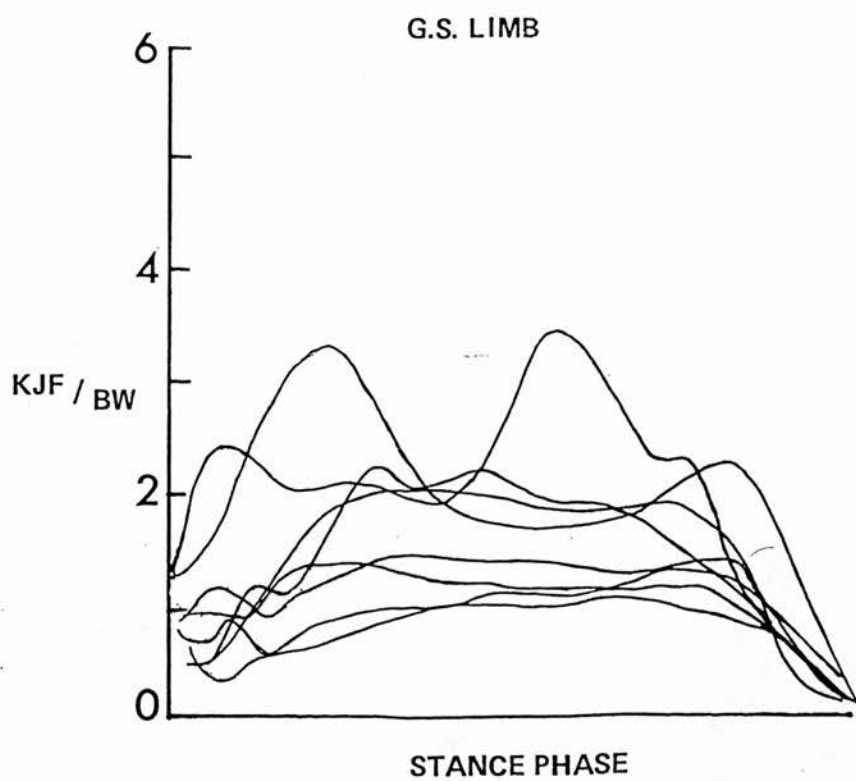


Fig. 4.59 Knee Joint Loads.

carried by the lateral joint compartment in the ipsilateral knees than in the contralateral knees, a situation more marked than that seen after total hip joint replacement ($25.25 \pm 21.42\%$ for the contralateral knees but $41.62 \pm 22.6\%$ for the ipsilateral knees).

Patient	RANGE OF MOVEMENT (Degrees)				Flexion	Abduction	Adduction	External Rotation	Internal Rotation	Leg Length Discrepancy cm.	Shoe Raise cm.	MUSCLE POWER* Abductors	Flexors	Walking Aid	Harris Score
17	0-100	0-15	0-20	10-25	-10					4.0	0.0	4	4	None	88
18	20-110	0-15	0-15	0-15	0-15					6.5	6.0	3+	3+	1stick	79
19	20-60	0-10	0-20	20-45	-20					6.0	1.5	3	3	None	74
20	15-45	0-15	0-20	0-30	0					4.0	1.5	2	3	1stick	61
21	0-100	0-30	0-20	0-30	0-30					3.0	2.0	3	3	1stick	72
22	10-50	-5	5	0-10	0					4.0	4.0	3	3+	1stick	74
23	0-70	0-15	0-10	10-20	-10					5.0	2.5	2	3	1stick	64
24	0-80	0-15	0-15	0-15	0					5.0	3.0	3	3+	None	73
25	0-70	0-20	0-30	20-40	-20					5.0	3.5	3	3	1stick	61

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- NOTE: 1. All Patients had a Positive Trendelenburg Test.
 2. Only Patient 25 scored less than 40 on the Harris Assessment.
 * M.R.C. Grading (Seddon 1954).

CHAPTER 4. TABLE 4.4 BIOMECHANICAL RESULTS - GIRDLESTONE EXCISION ARTHROPLASTY GROUP

Patient	Stance Phase Durn. (sec)	Muscle Forces			HJF (max) xBW	Phase H.S. % T.O. Stance	KJF (max) xBW	% Lat. Compartment Load		
		Abd.	Ext.	Flex. xBW	Quads					
17	GS	1.4	0.8	0.6	0.0	2.8	24	68	2.3	40
	H	2.7	0.0	1.8	1.1	4.1	20	88	3.0	23
18	GS	1.0	2.8	0.0	2.2	4.3	24	78	2.9	39
	H	2.7	2.3	1.8	1.9	5.0	27	70	2.9	43
19	GS	2.4	0.3	0.4	0.7	3.2	15	50	1.5	50
	H	1.8	2.2	0.5	1.9	3.3	24	55	2.0	50
20	GS	1.5	0.0	2.4	0.5	4.3	40	70	2.0	62
	H	2.3	1.2	0.7	1.2	4.8	50	50	2.5	23
21	GS	0.0	0.2	0.7	0.5	1.3	32	74	1.2	60
	H	2.4	1.4	0.2	1.1	3.9	25	72	2.1	43
22	GS	1.3	0.5	2.1	0.4	2.8	22	73	1.4	38
	H	1.9	2.0	1.2	1.6	4.2	16	72	2.6	29
23	GS	0.8	0.2	0.9	0.7	2.2	38	78	1.3	44
	H	3.1	0.8	1.1	1.1	4.8	24	76	2.5	11
24	GS	1.2	0.5	1.6	0.3	2.7	34	72	1.8	48
	H	1.8	0.8	1.5	2.2	3.6	26	78	3.0	15
25	GS	1.0	0.3	1.7	0.9	2.6	26	76	1.6	60
	H	1.7	1.2	1.5	1.7	3.8	23	72	2.7	38

4.V THE EFFECT OF ANATOMICAL VARIABLES UPON JOINT FUNCTION

Many total hip joint replacement prostheses are available for use. They may differ in one or more of several features, including stem configuration, head size and neck-stem angle. Some of these differences might be expected to influence the mechanics of the joint and, therefore, its function. When studying the influence of hip arthroplasty upon gait - a manifestation of joint function - the type of prosthesis used should be considered.

The variability in pattern of joint disease (vide infra) and the surgical technique used can influence the positioning of the components of the arthroplasty which will further affect the joint mechanics and function. Clearly, this contribution to joint function must not be ignored when studying gait or indeed any factor relevant to or resulting from hip arthroplasty.

1. Anatomical Variables

Because this study involves patients with only one hip involved all variations in hip joint anatomy have been assessed relative to the healthy hip joint. Measurements have been made from x-rays as described in Chapter 3.

The features which may change as a result of disease or surgery are :-

- i. The position of the hip joint centre relative to the pelvis which is determined by the position of the acetabulum and the shape of the femoral head.

- ii. The relationship of the centre of rotation of the hip joint to the greater and lesser trochanters which depends upon the neck-shaft angle, the neck length,

and the position of the greater trochanter.

Alterations in these parameters produce changes in muscle lengths and lever arms and will therefore be expected to produce changes in muscle forces and joint loads.

a. The Degenerative Hip Joint

Pre-operatively the major changes will be in the acetabulum and femoral head. The osteoarthritic joint commonly shows superior erosion of the acetabulum with or without superolateral subluxation and collapse of the femoral head (Solomon, 1976, Wroblewski and Charnley, 1982). The method of coding used to describe the pathological anatomy in the patients studied is shown in Figure 4.60 and Table 4.5 shows the distribution of these patterns. Wroblewski (1980) has also drawn attention to the antero-superior wear of the acetabulum but, as mentioned in Chapter 3, this has not been allowed for in the biomechanical analysis used in this work.

Figure 4.61 shows the relationship of the hip joint centres, measured with regard to the pelvis, to the position of the healthy hip joint centre. This Figure also indicates which patients were later treated by Charnley or Muller arthroplasties (vide infra) and shows that the two groups were dissimilar, those treated with the Charnley arthroplasty having the more severe anatomical change, pre-operatively.

b. The Replaced Hip Joint

Post-operatively the choice of prosthesis and its operative technique together with the positioning of its femoral and pelvic components will produce the reconstructed anatomy.

Acetabulum

A0	Normal anatomical appearance
A1	Early loss of superolateral margin
A2	Superioralisation of acetabular roof
A3	Severe destruction of acetabular roof
M0	No medialisation of socket
M1	Medialisation of socket to ilio- ischial line
M2	Medialisation beyond ilio-ischial line

Femoral Head

H0	Normal femoral head
H1	Flattening of femoral head
H2	Severe flattening of femoral head
H3	Total destruction of femoral head

Fig. 4.60 Radiological Grading of Hip Disease

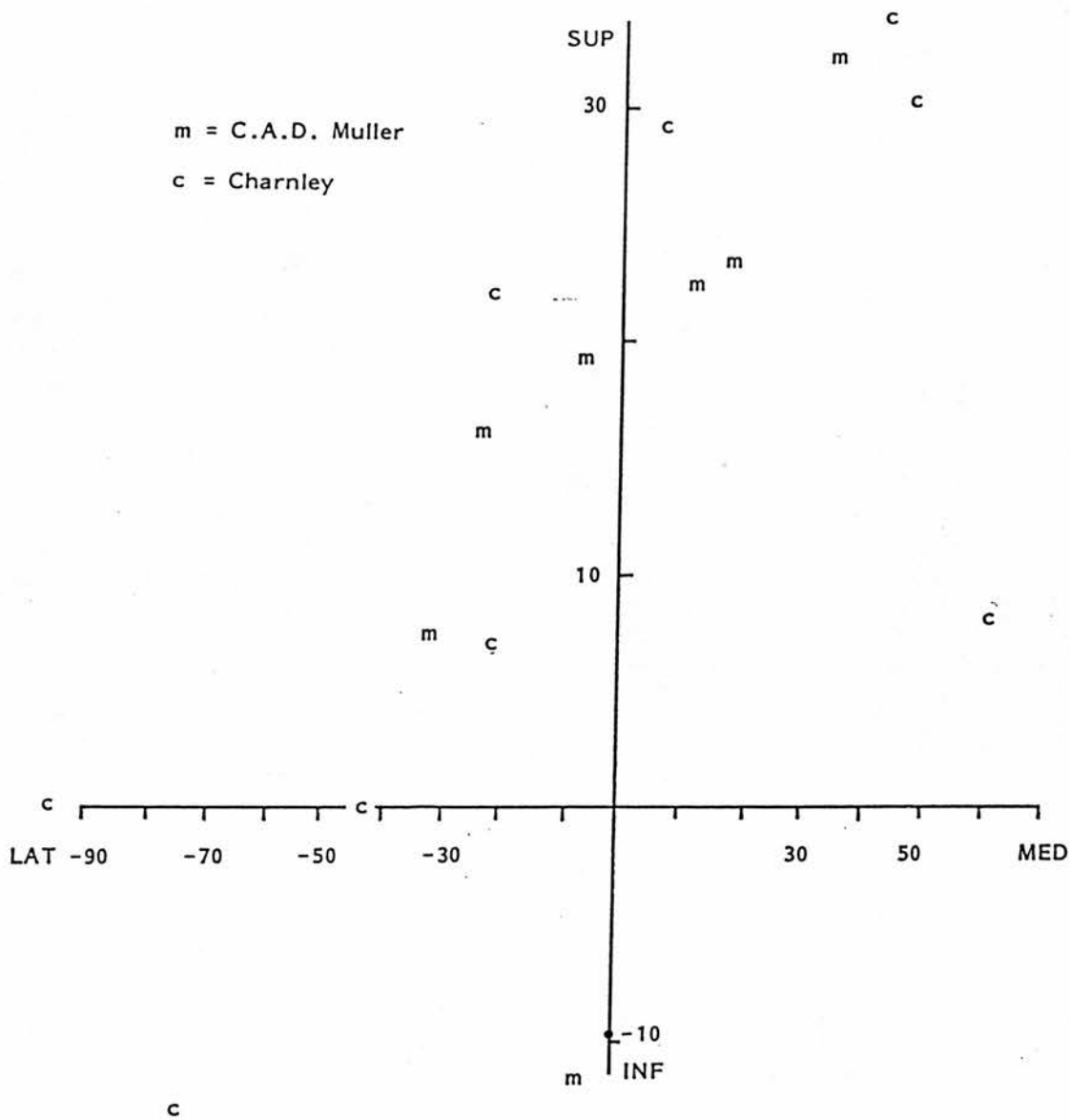


Fig. 4.61 Percentage Difference in Pelvic
 Position of Hip Joint Centres -
 Pre-Operative.

TABLE 4.5

X-RAY APPEARANCES OF AFFECTED HIP

<u>Patient</u>	<u>Diagnosis</u>	<u>Acetabulum</u>	<u>Head</u>	
			<u>Vertical</u>	<u>Horizontal</u>
1	O.A.	1	1	0
2	Congenital Coxa Vara	2	2	0
3	Mixed Connective Tissue Syndrome	1	1	0
4	Thompson Arthroplasty	1+	N/A	2
5	O.A.	1+	1+	0
6	Post- Traumatic Avascular Necrosis	1	1	2
7	C.A.D. Endo- Prosthesis	1+	N/A	0
8	O.A.	1	0	0
9	O.A.	1+	1	2
10	O.A.	3+	2	0
11	O.A.	1	0	0
12	O.A.	1	0	0
13	O.A.	1	1	0
14	O.A.	2	1	0
15	O.A.	1+	1	0
16	O.A. Old C.D.H.	3	2	0

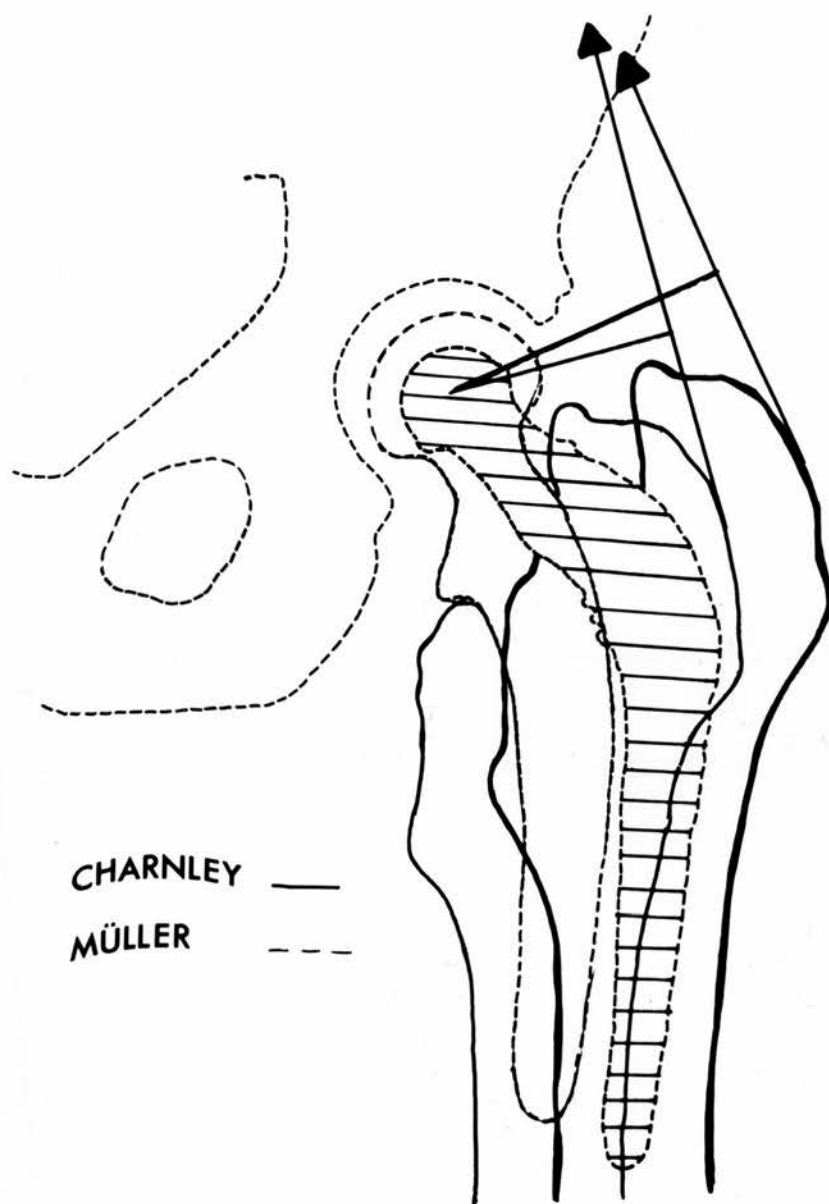
The two designs used were the Charnley (Thackray) and C.A.D. Müller (Howmedica) hip replacements (Fig. 4.62). The Charnley device used had a 22 mm diameter head, a 40 mm offset, a stem neck angle of 130° , a choice of neck lengths (with an appropriate change in neck-stem angle to maintain a 40 mm offset) and a variety of stem configurations. The C.A.D. Müller femoral component had a 32 mm diameter head, a neck-stem angle of 147° and a choice of neck lengths (again with differing neck-stem angles to maintain a constant offset). The stem was of I-beam construction and available in straight or curved forms, each in standard and small sizes. The C.A.D. Müller device will therefore produce a more valgus reconstructed hip joint (Fig. 4.63).

The method used for the insertion of these prostheses were alluded to in Chapter 2. The two major differences were in the surgical approach where trochanteric osteotomy and later re-attachment was always performed for the Charnley arthroplasty but never for the C.A.D. Müller, and in the preparation of the acetabular cavity where standard reamers were used with the Charnley and gouges were used with the C.A.D. Müller. These factors meant that with the Charnley arthroplasty the greater trochanter could be re-attached in a displaced position, thus influencing the abductor muscle lengths and lever arms, and the acetabular cavity could be deepened more efficiently again affecting the mechanics of the joint.

It can be seen from Figure 4.64 that the position of the reconstructed hip joint centre relative to the pelvis was nearly always more medial and more superior than that



Fig. 4.62 Charnley and C.A.D. Müller
Femoral Prostheses.



4.63 Diagram to illustrate the effects of using the Charnley and C.A.D. Müller prostheses upon femoral offset and abductor muscle lever arm.

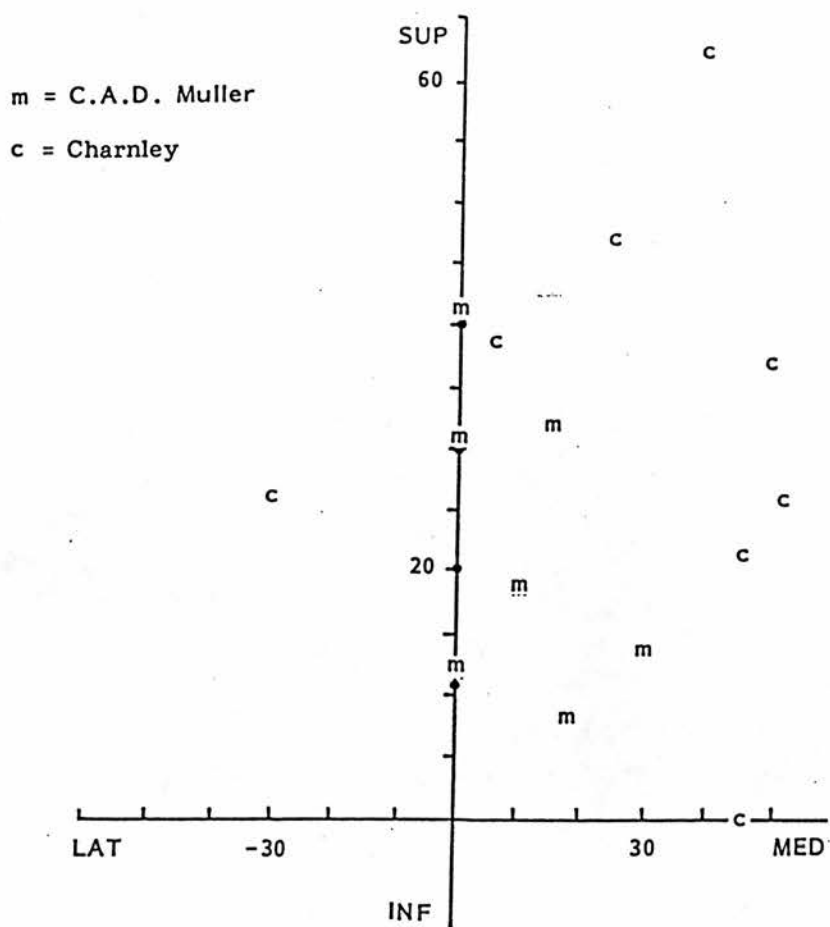


Fig. 4.64 Percentage Difference in Pelvic
Position of Hip Joint Centres -
Post-Operative.

of the healthy hip and that the more extreme positions were achieved by the use of the Charnley arthroplasty. This is only one factor which may contribute to the pattern of hip joint loads described in the first part of this Chapter and the other anatomical variables must be considered.

Additionally, the results already presented show that the loading at the knee joints is also altered both in hip disease and after arthroplasty and the morphological changes at the hip joint must be considered in relation to these parameters.

2. Statistical Analysis

a. Multiple Linear Regression Analysis

Because of the number of anatomical variables which could influence the joint loadings and the need to consider both knees as well as both hip joints, the contribution of each variable to the variance in joint load was studied using multiple linear regression analysis using the I.C.L. 2980 Computer at the University of Strathclyde and the Statistical Package for the Social Sciences Computer software (S.P.S.S.). Colinearity amongst the variables was checked by generating a Pearson correlation matrix. Additionally multiple correlations were performed and plotted, the value of these being limited by the multi-factorial nature of the problem under investigation.

The independent variables were introduced into the regression equations in a predetermined step-wise fashion in an attempt to mimic the steps taken during surgery or the process of degeneration.

The order was :-

- i. (Post-operatively only) the prosthesis chosen.
- ii. The vertical position of the hip joint centre with respect to the pelvis.
- iii. The medial position of the hip joint centre with respect to the pelvis.
- iv. The neck lengths taken as the vertical distance between the lesser trochanter and the hip joint centre.
- v. The position of the greater trochanter, measured as its lateral displacement from the femoral axis.

Tables 4.6a and 4.6b give the values for R squared change for the entire pre-operative and post-operative groups. This value indicates the contribution made by the named variable to the variance of the load at the joint indicated. It should be noted that since variables were introduced in a step-wise fashion even small values for R squared change for a variable introduced late in the chain may be important particularly if the preceding variables have explained the large percentage of the variance. Tables 4.7 and 4.8 give the values of the same parameter for the C.A.D. Müller and Charnley groups respectively.

All groups studied are of small numbers and this makes the interpretation of the data difficult since if a given variable is important in only two to three cases of a small group the results might suggest that it is important to the group as a whole. For this reason it is difficult to place much reliance on conclusions drawn from this data.

CONTRIBUTION OF ANATOMICAL VARIABLES TO VARIANCE IN JOINT LOADS

TABLE 4.6a Hip Disease Group n = 14

Joint	Anatomical Variable	R ² Change (as %)
Healthy Hip	Yc	9.3
	Zc	9.0
	N.L.	19.8
Affected Hip	Yc	37.0
	N.L.	9.1
Contralateral Knee	Yc	4.4
	N.L.	23.3
Ipsilateral Knee	Yc	26.9
	Zc	9.4

TABLE 4.6b Hip Joint Replacement Group n = 16

Joint	Anatomical Variable	R ² Change (as %)
Healthy Hip	F.D.	17.6
	N.L.	5.1
Replaced Hip	F.D.	29.1
	Yc	11.0
Contralateral Knee	Yc	13.4
	G.T.Z.	21.3
Ipsilateral Knee	Yc	4.4
	G.T.Z.	4.8

CONTRIBUTION OF ANATOMICAL VARIABLES TO VARIANCE IN JOINT LOADS

TABLE 4.7 C.A.D. Muller Group n = 7

Joint	Anatomical Variable	R ² Change (as %)
Healthy Hip	N.L.	32.4
	G.T.Z.	16.3
Replaced Hip	Yc	20.9
	Zc	33.6
Contralateral Knee	Yc	11.2
	N.L.	14.2
	G.T.	42.3
Ipsilateral Knee	Zc	48.8
	G.T.Z.	35.0

TABLE 4.8 Charnley Group n = 9

Joint	Anatomical Variable	R ² Change (as %)
Healthy Hip	Yc	8.0
	Zc	11.7
Replaced Hip	Yc	41.7
	Zc	4.4
	N.L.	23.5
Contralateral Knee	Yc	32.3
	Zc	5.2
Ipsilateral Knee	N.L.	26.9
	G.T.Z.	19.5

Pre-operatively, the most notable relationship appears to be between the vertical position of the hip joint centre and the load across the affected hip and the ipsilateral knee joint. The neck length of the affected hip also appears to have an important influence across that joint but even more effect upon the load across the healthy hip joint and contralateral knee joint.

Post-operatively, the load across the replaced hip joint is influenced most by the design of the femoral component used (i.e. the type of hip arthroplasty performed) but it is also strongly influenced by the vertical position of the acetabular cup. This variable also has a notable influence upon the loads at the contralateral knee but at this joint the lateral position of the greater trochanter would appear to be the most important variable.

When one considers the C.A.D. Müller and Charnley groups separately it is important to note the very small numbers involved. However in the Müller group over 50 per cent of the variance in the load at the replaced hip joint is explained by the vertical and horizontal positioning of the acetabular cup whereas in excess of 80 per cent of the variance in the load at the ipsilateral knee is explained by the horizontal position of the cup and the lateral position of the greater trochanter. This last variable also makes a great contribution to explaining variance in loads at the contralateral knee joint.

In the Charnley group over 60 per cent of the variance in the load at the replaced hip joint is explained by the vertical position of the cup and the length of the

neck and this last variable is also important with regard to the load at the ipsilateral knee, together with the lateral position of the greater trochanter.

The design of the prosthesis would appear to have virtually no influence upon the loads across the knee joints.

b. Correlations

Despite the obvious limitations of correlations in this multifactorial situation several interesting findings can be presented. Pre-operatively Figure 4.65 shows the strong correlation between the vertical position of the joint centre and the mean load at the healthy hip ($p = 0.05$) and Figure 4.66 shows the even stronger correlation for the load at the diseased hip ($p = 0.002$). Similar strong correlations occur between the Y position of the joint centre and the loads at the ipsilateral ($p = 0.008$) and contralateral knees ($p = 0.06$), (Figs. 4.67 and 4.68).

The medio-lateral position of the joint centre also correlates strongly with the loads across the diseased and healthy hip joints (Figs. 4.69 and 4.70). ($p = 0.003$ and $p = 0.003$). At the knee joints the correlation is not strong.

Neck length correlated only poorly with all the joint loads considered.

Post-operatively correlations have been considered for the Charnley and Müller groups separately.

For the small Müller group, correlations are seen between the medio-lateral position of the cup and the joint

$$\frac{Y(\text{AFF}) - Y(\text{HEALTHY})}{Y(\text{HEALTHY})} \%$$

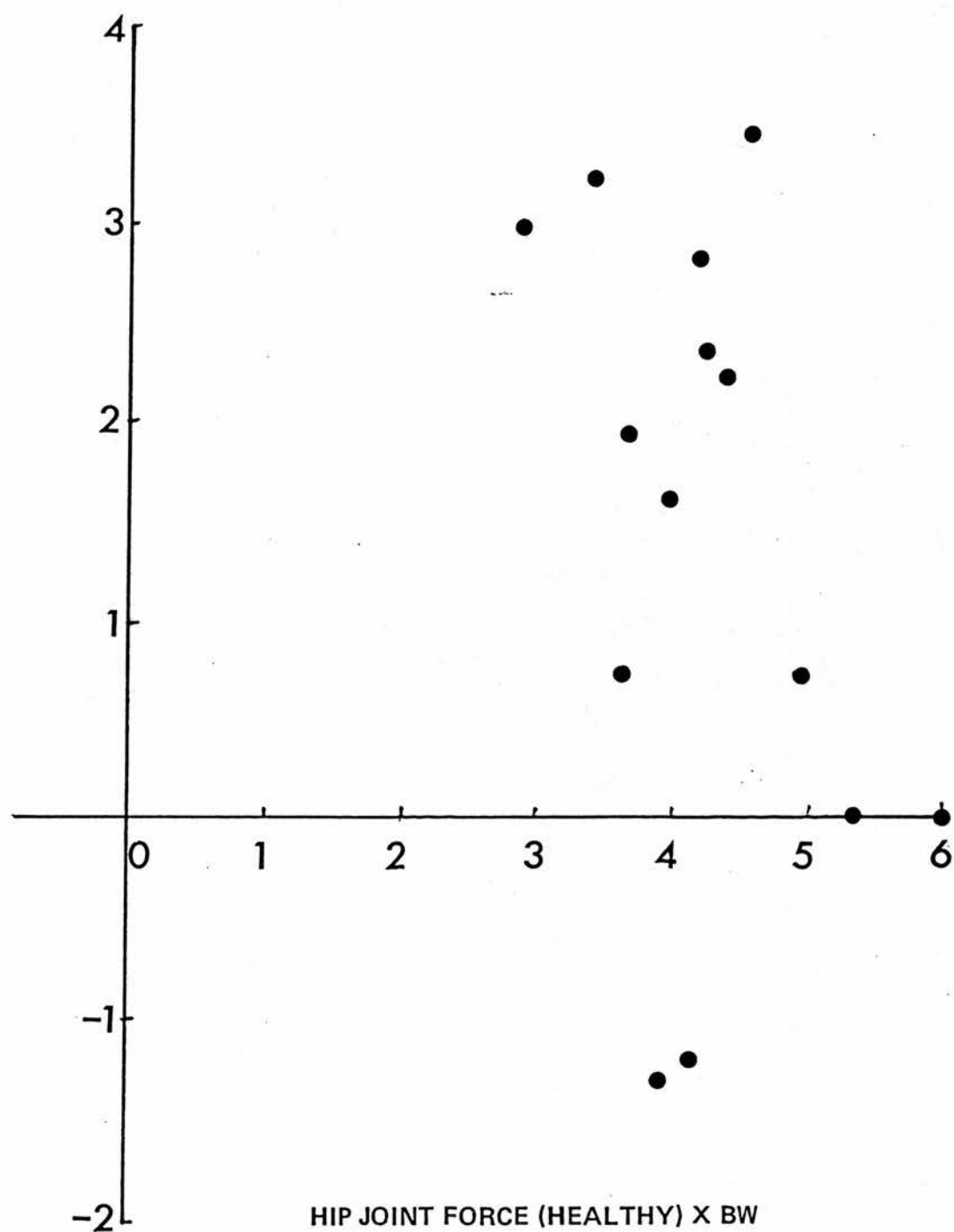


Fig. 4.65 Correlation of vertical pelvic position of hip joint centre with healthy hip joint load.

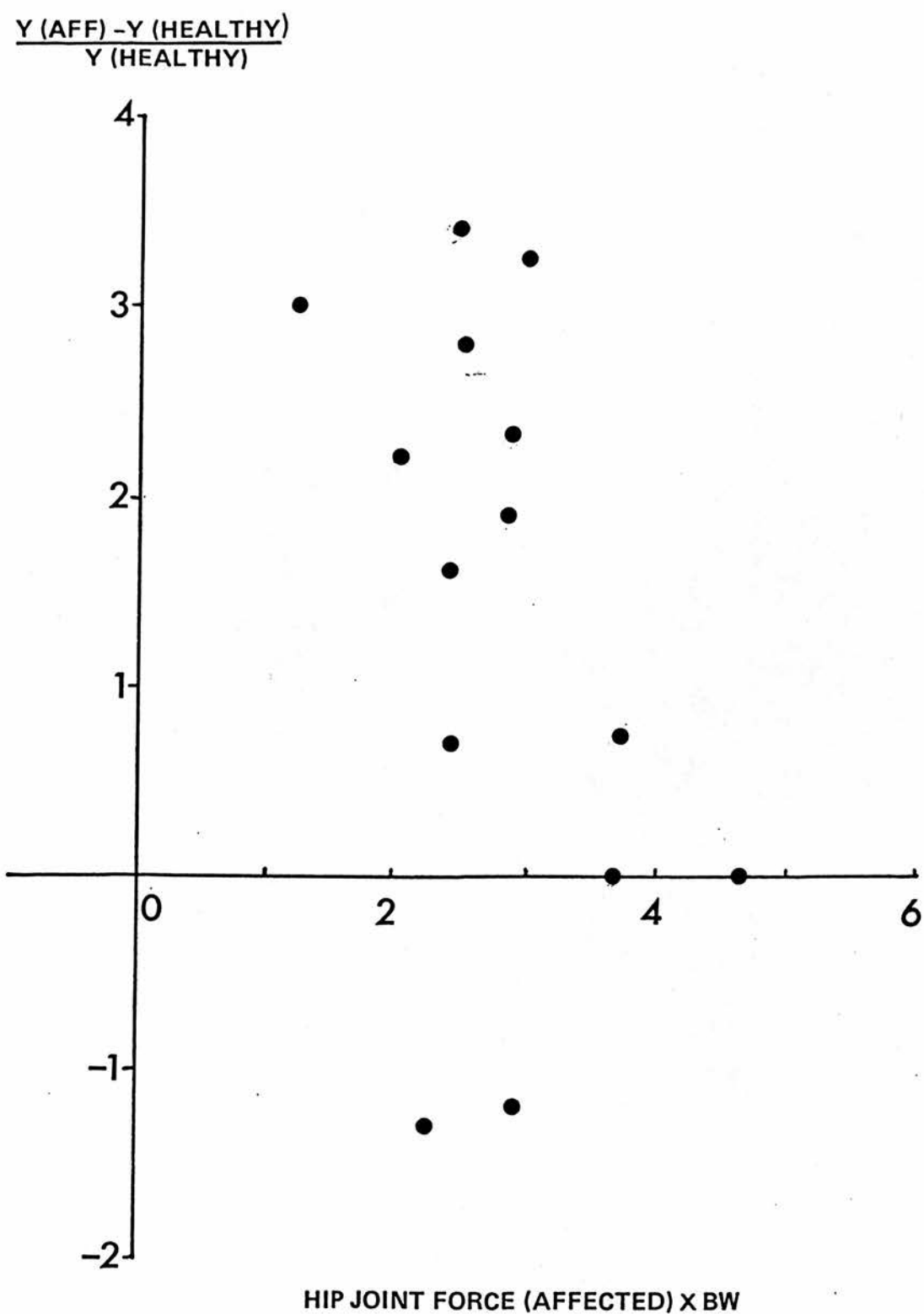


Fig. 4.66 Correlation of vertical pelvic position of hip joint centre with affected hip joint load.

$$\frac{Y(\text{AFF}) - Y(\text{HEALTHY})}{Y(\text{HEALTHY})} \%$$

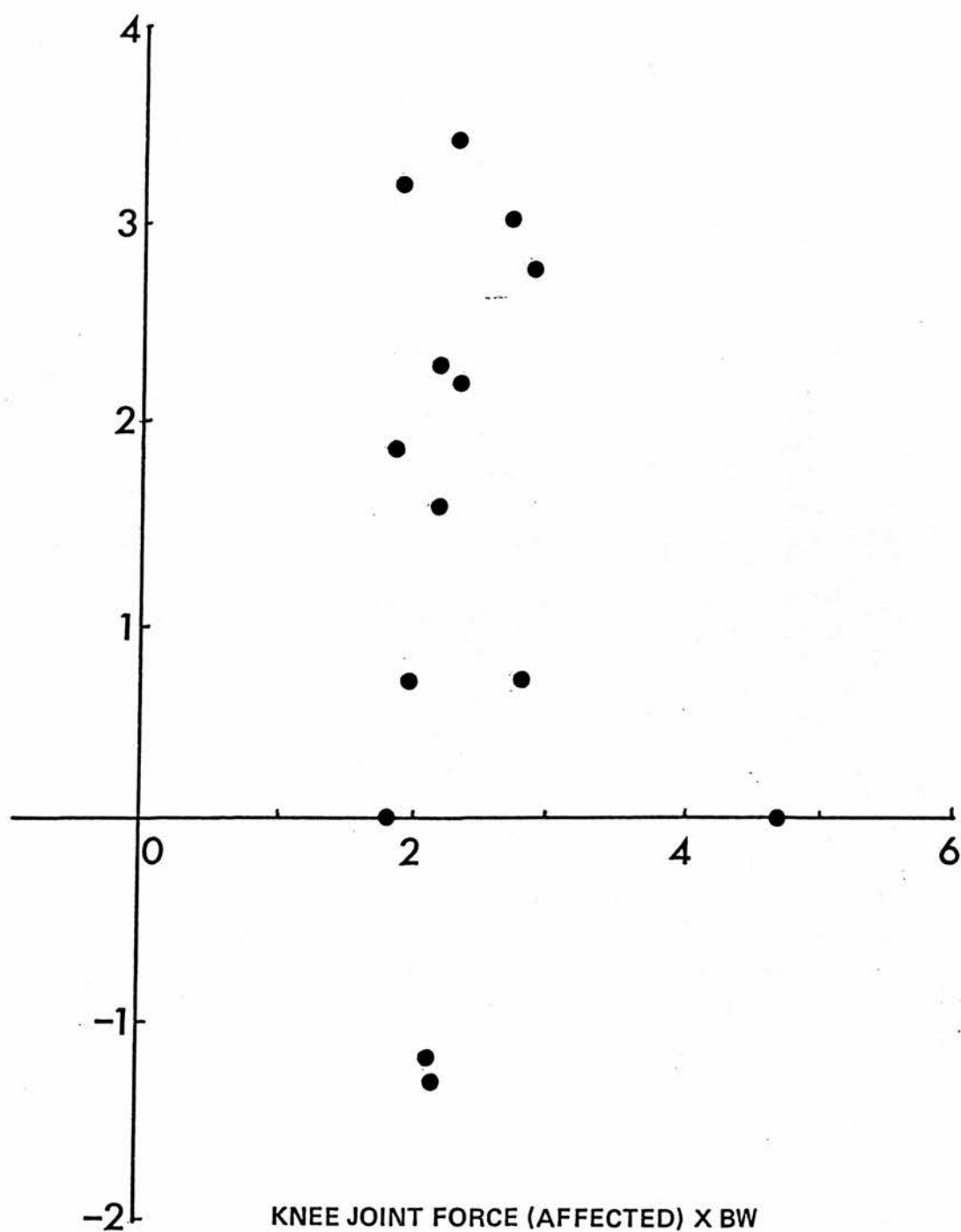


Fig. 4.67 Correlation of vertical pelvic position of hip joint centre with ipsilateral knee joint force.

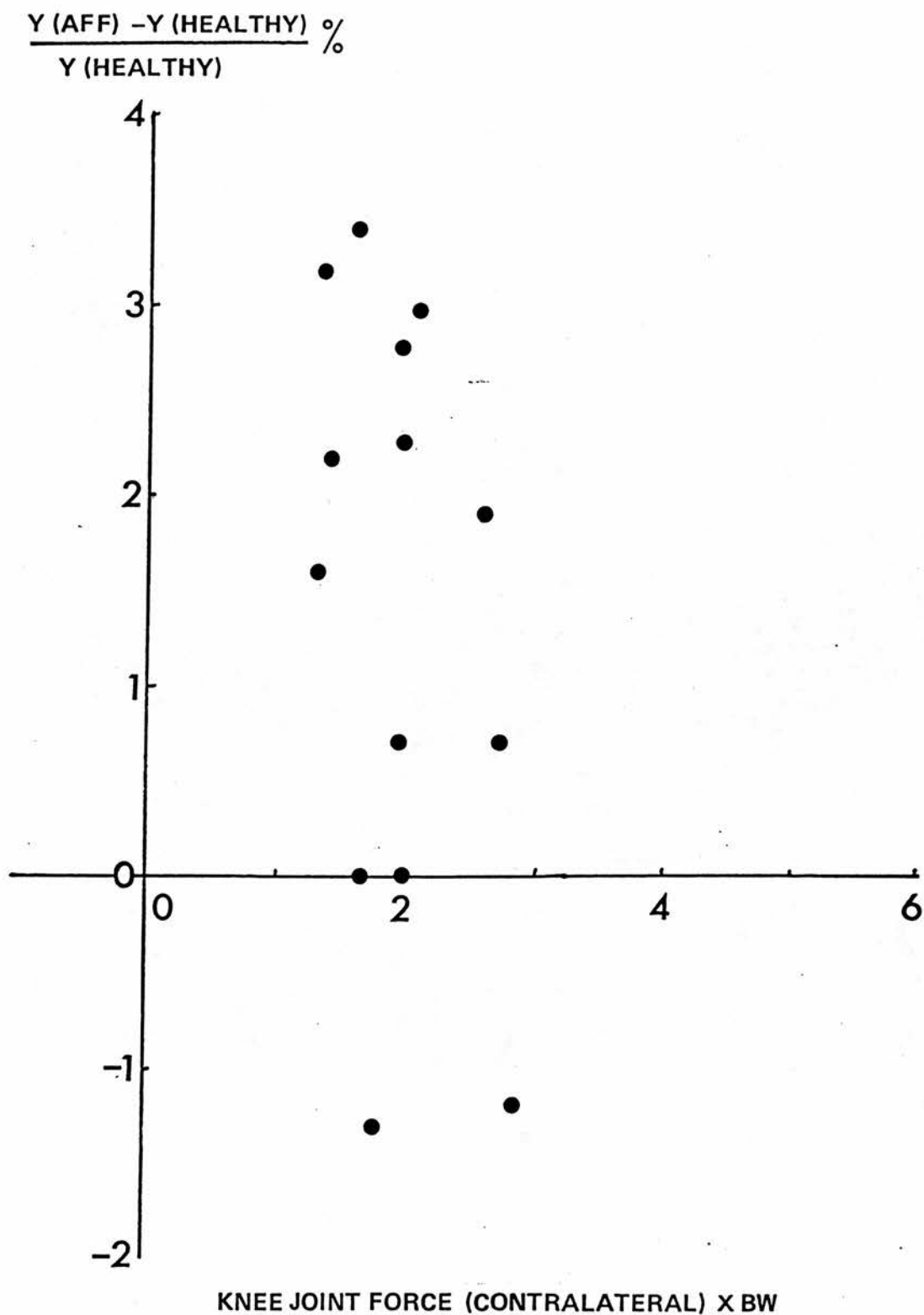


Fig. 4.68 Correlation of vertical pelvic position of hip joint centre with contralateral knee joint force.

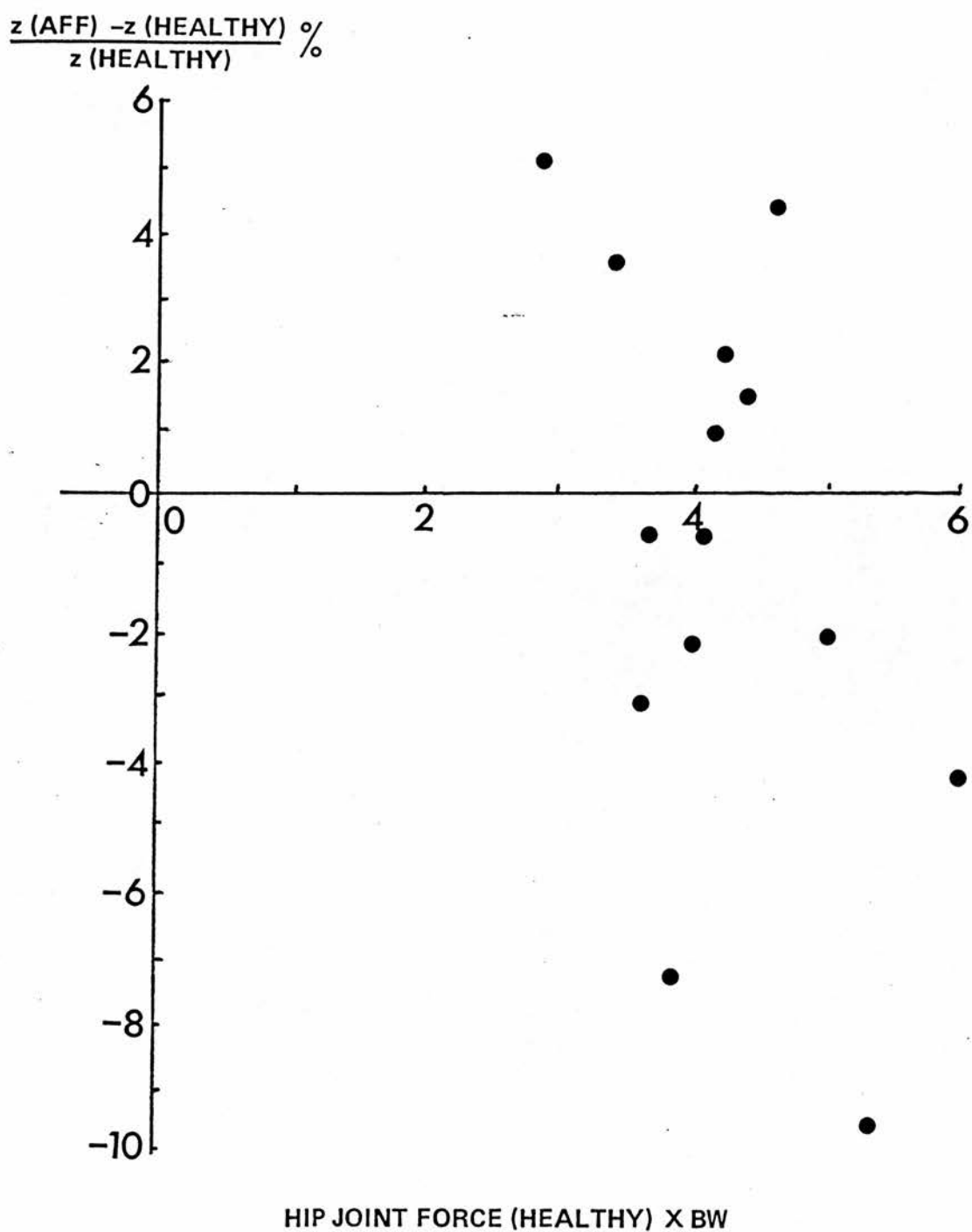


Fig. 4.69 Correlation of horizontal pelvic position of hip joint centre with healthy hip joint force.

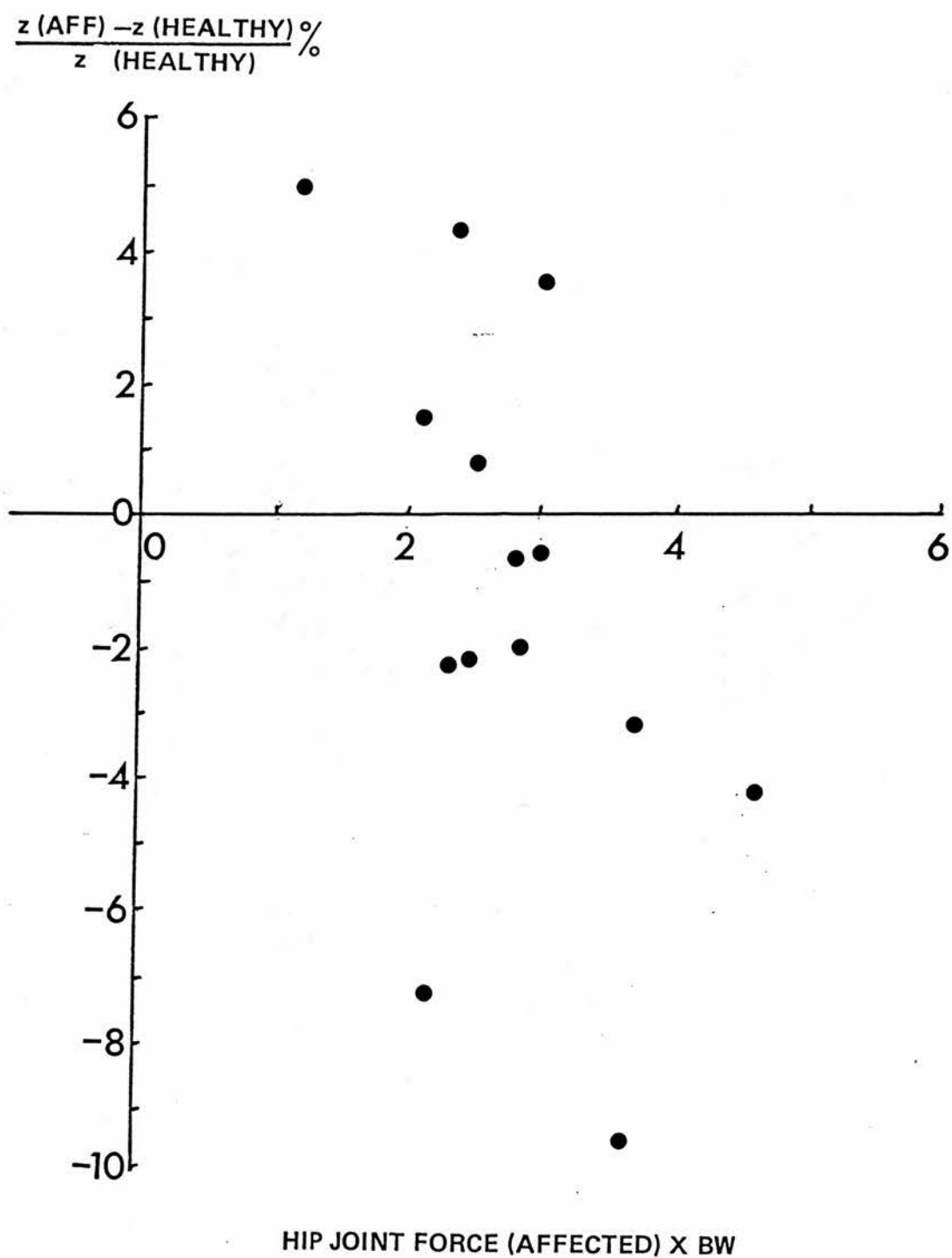


Fig. 4.70 Correlation of horizontal pelvic position of hip joint centre with affected hip joint force.

load across the replaced hip and ipsilateral knee ($p = 0.07$ and $p = 0.05$). No other variable achieved even the 90 per cent significance level.

For the similarly small Charnley group, no variable correlates at even the 90 per cent significance level with the loads across the hips or knees.

TABLE 4.9 MEASUREMENTS FROM X-RAYS

H - Healthy A - Affected R - Replaced Hip Joint

PATIENT		Y/a	Z/a	$G.T/a$	$N.L./a$	$Labd/a$
1	H	0.23	0.10	0.23	0.27	0.19
	A	0.21	0.12	0.22	0.26	0.16
	R	0.22	0.12	0.19	0.25	0.19
2	H	0.21	0.14	0.17	0.20	0.20
	A	0.23	0.08	0.13	0.20	0.25
	R	0.19	0.13	0.19	0.20	0.19
3	H	0.25	0.14	0.20	0.23	0.20
	A	0.24	0.11	0.23	0.21	0.20
	R	0.25	0.14	0.14	0.23	0.15
4	H	0.26	0.14	0.21	0.23	0.21
	A	0.25	0.20	0.20	0.24	0.24
	R	0.25	0.17	0.22	0.18	0.27
5	H	0.21	0.16	0.23	0.25	0.18
	A	0.20	0.13	0.24	0.23	0.18
	R	0.16	0.16	0.14	0.26	0.17
6	H	0.20	0.20	0.21	0.20	0.22
	A	0.19	0.22	0.18	0.20	0.21
	R	0.19	0.22	0.22	0.22	0.23
7	H	0.28	0.16	0.23	0.24	0.23
	A	0.26	0.13	0.20	0.23	0.23
	R	0.27	0.16	0.24	0.24	0.24
8	H	0.29	0.14	0.21	0.19	0.21
	A	0.29	0.14	0.19	0.18	0.20
	R	0.28	0.17	0.14	0.19	0.19
9	H	0.34	0.14	0.18	0.20	0.19
	A	0.31	0.18	0.21	0.20	0.24
	R	0.28	0.20	0.19	0.17	0.23

TABLE 4.9 continued. MEASUREMENTS FROM X-RAYS

H - Healthy A - Affected R - Replaced Hip Joint

PATIENT		y/a	z/a	$G.T/a$	$N.L./a$	$Labd/a$
10	H	0.28	0.10	0.18	0.21	0.20
	A	0.28	0.00	0.13	0.15	0.17
	R	0.20	0.14	0.22	0.24	0.21
11	H	0.19	0.14	0.20	0.21	0.20
	A	0.17	0.13	0.20	0.21	0.27
12	H	0.22	0.12	0.20	0.19	0.20
	A	0.19	0.15	0.19	0.18	0.16
	R	0.20	0.15	0.13	0.20	0.16
13	H	0.24	0.09	0.19	0.19	0.20
	A	0.23	0.07	0.21	0.19	0.19
	R	0.22	0.09	0.19	0.29	0.16
14	H	0.24	0.13	0.22	0.21	0.22
	A	0.21	0.12	0.21	0.21	0.19
	R	0.22	0.15	0.20	0.17	0.21
15	H	0.25	0.13	0.18	0.20	0.20
	A	0.22	0.12	0.19	0.20	0.20
	R	0.24	0.16	0.14	0.22	0.14
16	H	0.24	0.10	0.22	0.22	0.19
	A	0.24	0.06	0.14	0.09	0.21
	R	0.24	0.15	0.20	0.19	0.23
25	H	0.25	0.14	0.26	0.26	0.24
	R	0.15	0.14	0.24	0.19	0.20

4.VI SUMMARY OF RESULTS

1. Hip Disease Group

The pattern of gait was abnormal in these patients with a slow walking velocity, a reduced period of stance on the affected limb, and a typical pattern of hip and knee movements in each limb. The pattern of loading at the diseased hip joints was also abnormal, the resultant moments and muscle forces suggesting abductor muscle impairment as the main cause.

The magnitude of the loads transmitted by the affected hip joint were much reduced when compared with those in normal subjects. The loads at the healthy hip joints were similar in magnitude to those in the normal subject and exhibited a similar pattern.

Abnormalities of loading were also seen at the knee joints with an abnormal pattern at the ipsilateral knee.

It was not possible to correlate clinical measurements of function or movement with the pattern or magnitude of joint loads.

2. Hip Joint Replacement Group

Abnormalities of the gait pattern persisted after total hip replacement but were less marked than in the pre-operative group with an increased walking velocity and symmetrical stance phase duration (i.e. the antalgic element of the gait was lost).

An incomplete recovery in the pattern of loading at the replaced hip joint was seen, the appearances again suggesting abductor and/or extensor muscle dysfunction; a suggestion further validated by consideration of the

resultant moments and muscle forces. Loads at the healthy hip joints were at the upper end of the normal range and the pattern was as seen in normal subjects. Differences were noted for the two hip replacement devices used, a higher load being transmitted across the Muller hip.

At the knees there was an incomplete return towards the normal pattern of movement in the operated limbs with a lower level of loading than that seen at the contralateral knee. More of the load at the ipsilateral knee was transmitted through the lateral compartment but this was subject to great variability for which no cause was evident from the information available.

The range of movement used at the hip and knee joints during level walking was less than that available for use as indicated by clinical examination.

3. Girdlestone Excision Arthroplasty Group

This heterogeneous group of patients exhibited a disturbance of gait pattern similar to but more severe than that seen in the other two groups.

Severe dysfunction of the hip joint musculature, especially the abductor muscles, is again suggested by the resultant moment and muscle force data and this was reflected in the patterns of loading at the GS hip. The loading pattern was also abnormal at the healthy hip where considerably greater loads were transmitted.

At the knee joints, the loads were similar in magnitude but different in pattern, with a high proportion of load being transmitted by the lateral compartment of the ipsilateral knee.

4. Anatomical Variables and Hip Joint Function

Pre-operatively the more superiorly placed was the centre of rotation of the hip joint i.e. the more destroyed was the joint, the lower was the load transmitted by that joint and the contralateral hip joint. Similarly, the shorter was the neck length the lower was the load across the affected hip but the higher were the loads across the healthy hip joint and the contralateral knee joint.

Post-operatively the reconstructed hip joint centre was nearly always superior and medial to the position suggested by consideration of the healthy hip joint.

The load across the replaced hip joint is most strongly influenced by the type of arthroplasty performed but is also affected by the vertical position of the acetabular cup in the pelvis and by the lateral position of the greater trochanter.

The magnitude of loads across the knee joints appears to be associated with the position of the acetabular cup, the neck length and the lateral position of the greater trochanter but the small numbers involved make meaningful conclusions difficult to establish.

The type of hip arthroplasty performed has no influence upon the knee joint loads.

CHAPTER 5

DISCUSSION

I INTRODUCTION

II HIP DISEASE GROUP

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2. Hip Joints
3. Knee Joints
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1. Loads and Movements at the
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2. Anatomical Variables
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5.I INTRODUCTION

The three groups of patients considered in this thesis exhibit variations from normal in the anatomy of one hip joint. Since the laws of mechanics apply to the human frame this changed anatomy must be reflected in changes in the mechanics at that joint. However, it must not be forgotten that in this biological system bone and muscle will adapt to altered mechanics in time and the force of muscular contraction depends not only upon mechanics but also upon neural input and such abstract factors as acquired patterns of movement and pain. This last factor complicates the interpretation of data from the degenerative hip disease group, but is not a feature of the other two groups.

In all groups of patients studied the suggestions of McGrouther (1975) and Tooth (1976) that abnormalities in one load bearing joint of the lower limbs affect all the other load bearing joints has been substantiated.

The aim of this final Chapter will be to examine the results for each of the groups of patients critically in the light of earlier work, establishing conclusions where possible and making suggests for the elucidation of outstanding problems.

5.II HIP DISEASE GROUP

1. Gait Pattern

This was a group in which there was a wide range of anatomical variation at the hip joint and in which every patient was suffering pain which imposed restrictions upon their mobility and lifestyle.

In common with Murray and her colleagues (1971) and Smidt and Wadsworth (1973) it has been found that this group of patients had a slower walking speed than the normal subjects and the inequality of successive step lengths was noted but not measured. Additionally the shorter duration of stance phase on the affected limb was noted.

Further, in common with other authors (Murray et al, 1971; Stauffer, Smidt and Wadsworth, 1974; and McGrouther, 1975), the asymmetry of the gait in these patients has been confirmed with a smaller flexion extension range being used at the involved hip joint than at the sound hip joint, this being mainly due to the limitation of hip extension in the later part of stance phase. As the above authors had also noted the range of flexion and extension used during level walking was considerably less than that found to be present upon clinical examination.

McGrouther's findings (1975) of altered hip moments in both lower limbs, particularly the abductor moments, have been confirmed and, as will be mentioned below, the results of this study are also in agreement with McGrouther with regard to the hip and knee joint loads.

2. Hip Joints

There was a great variation in the pattern of hip joint load and, therefore, in the pattern of muscular activity. However, despite this variation in pattern, the peak loads at the diseased hip joints were within the bounds of hip joint loads reported by Paul (1967) (I Fig.1.18) for normal subjects, although lower than those transmitted at the contralateral healthy hip joints. Since pain is a major feature in degenerative hip joint disease it would seem reasonable to suggest that each subject will adopt a gait which will minimise pain. Clearly, from the results presented, although there is a reduction in the load passing across the diseased hip joint a considerable force is being transmitted and the so-called "antalgic" gait is not achieving pain relief by load reduction. As mentioned above, in these patients the walking velocity is small, the duration of stance phase on the affected limb is reduced, thus reducing the duration of load bearing, and movements at the hips and knees of the affected limbs are markedly restricted. McGrouther (1975) has suggested that the pain is related to joint movement and occurs at the extremes of the range of movement. He has hypothesized that the restricted movement is the major pain avoidance mechanism during walking. The results presented in this thesis are similar to those of McGrouther but show more reduction of the load at the diseased hip joints than in his series. Reduced movements at joints result in the need for lower muscle forces and therefore the need to lower joint loads. Hence

McGrouther's hypothesis requires the finding of reduced joint loading.

Does the gait used by these patients result in pain relief? It would appear that it does not since all the patients reported pain interfering with their daily activities and walking distance despite the type of gait they employed. The origin of the pain in the osteoarthritic hip joint is not clear. It may be a result of stretching of the thickened capsule, or of loss of the shock absorbing capacity of the articular cartilage. Restriction of movement would not seem to be very successful in relieving pain which suggests that the state of the joint surfaces has some influence in this regard. Patients who use a walking stick use a greater range of movement than when not using a stick (Brand and Crowninshield, 1980). Such patients also reduce the load across their diseased hip joints by up to 60 per cent when using a stick, but transmit load well within the range seen in normal individuals when walking unaided. Subjectively they complain of less pain. This suggests that hip contact force may be responsible for the pain in the arthritic hip rather than the stretching of the capsule. In this study the reduction in hip contact force seen at the diseased hips compared to the healthy contralateral hips is modest and may not be sufficient to affect the perception of pain (healthy hip joint load 4.83 ± 2.04 xBW; diseased hip joint loads 2.98 ± 1.78 xBW).

It can be argued that painful stimuli from the hip joint will inhibit muscle action to such an extent that in order to maintain equilibrium at the joint, movements will

have to be restricted. This will reduce the magnitude of the moments required and leads to the hypothesis that the gait used by patients with degenerative hip disease is a result of the pain and not merely a means of relieving it. Some support is lent to this hypothesis by the clinical observation of the shortening of stride length and width with increasing severity of disease.

Additionally, if hip pain is felt only at the extremes of movement, as suggested by McGrouther (1975), reducing the range should result in painfree gait or pain at heel strike and/or toe-off. This might be expected to be reflected in abnormalities in the joint load or moment traces at these times in the gait cycle. However, as has been demonstrated in Chapter 4, abnormalities are seen throughout stance phase in these traces and this strongly suggests a continuous effect upon muscle action.

The suggestion that the reduction in muscular activity is attributable to pain and not merely to a change in joint mechanics secondary to the altered anatomy, requires justification. It has been shown (Solomon, 1976, Wroblewski and Charnley, 1982) that the usual configuration of the degenerative hip joint comprises supero-lateral migration of the femoral head with or without concomitant erosion of the acetabulum or femoral head. This was the pattern seen in this group of patients (Chapter 4, Table 4.5). The major effect upon the capacity of the muscles to generate the required moment is, therefore, a reduction in their resting fibre length, since there is little or no change

in the length of their lever arms. However, muscle has the capacity to adapt to shortening with time and would therefore be expected to be capable of generating a "normal" degree of force in the absence of other inhibitory factors.

The reduction in muscle forces and moments is seen to a similar degree in all the hip muscle groups (Table 4.2a) and this provides further support for the theory of pain inhibition of muscle action as the cause of the gait pattern.

3. Knee Joints

Another feature of this gait pattern is the difference between the two knee joints (Chapter 4 II 2b). The very much reduced movements at the ipsilateral knee are consistent with the need to reduce the requirement for muscle action at the hip joint. The increased flexion range at the contralateral knee would seem to be a compensatory mechanism for the asymmetrical step lengths which result from the differing ranges of hip movement. If the normal pattern of symmetrical knee movements were to be seen there would have to be more extension range at the diseased hip than is seen in these patients. Since all patients in this group had flexion contractures they were unable to extend the hip upon the pelvis. However, by increasing the lumbar lordosis it is possible to bring the limb posterior to the vertical. Of course, in the type of patient who requires hip joint surgery the lumbar spine may also be degenerate and have limited movements compromising the hip movements further. The increased post heel strike flexion at the contralateral healthy knee may be a result

of the hyperlordosis movement or may be a consequence of the need to lower the pelvis at the time of heel strike to accommodate the pre toe-off knee flexion on the affected side (see Chapter 5.III.1).

The difference in movement patterns between the knees results in differences between the moments required for equilibrium, and, since these joints are normal in anatomy, in the values of muscle force and joint load. Greater quadriceps action is required at the contralateral knee and consequently a greater knee joint load is seen.

The variability in the distribution of compartmental joint loading (Chapter 4.II.2f) may reflect the variability in trunk movements during gait, patients with large trunk excursions towards the supporting diseased limb having more load transmitted across the lateral compartments. Trunk movements were not measured in this study.

4. Conclusions

i. Patients with degenerative hip disease exhibit an asymmetrical gait pattern with a slow walking velocity, reduced duration of stance phase on the affected limb, a limited range of movements at the affected hip, reduced flexion at the ipsilateral knee with increased flexion at the contralateral knee.

ii. It is hypothesized that this gait is the result of pain rather than an attempt to relieve pain.

iii. The increased knee flexion seen at the contralateral knee is thought to be a mechanism for compensating for the painful limited extension at the diseased hip. It results in an increased load at the contralateral knee joint.

iv. The load at the diseased hip joint is reduced as a result of the reduced muscle action and decreases with the increasing severity of disease.

5.III HIP JOINT REPLACEMENT GROUP

1. Gait Pattern

Lack of pain is a conspicuous feature following successful hip joint replacement surgery and every patient in this study experienced this. However, significant abnormalities of gait persisted in all patients after total hip arthroplasty. This had been noted previously by several authors (Stauffer et al, 1974, Murray et al, 1975, Gore et al, 1977, Murray et al, 1979, 1981a, 1982, Jacobs, Skorecki and Charnley, 1972, Andriacchi et al, 1980). These patients exhibited a greater walking velocity than that seen pre-operatively although it was still slower than the velocity reported for normal subjects. There was no significant difference between the duration of stance phase in each limb and biomechanical testing was considerably easier by virtue of the more symmetrical step lengths. A greater range of movement was seen at the replaced hip than had been present at the diseased hip but the movements of the two hip joints were not symmetrical there being less excursion into extension on the replaced side. Again, the range of hip flexion and extension during walking was considerably less than that available.

As will be discussed further below, a most conspicuous feature in gait was a failure of recovery in the pattern of the abductor moment about the hip joint. Flexor and extensor moments remained abnormal but the changes were not as striking as those in the abductor-adductor group. The magnitude of these moments was approximately the same for both limbs.

Unlike Andriacchi and his colleagues (1980) I found no correlation between the length of the abductor lever arms and the flexion-extension range used at the replaced hip joint. However, like them, I found no correlation between the length of the abductor lever arm and the abduction/adduction moments (MHX).

2. Hip Joints

The peak hip joint loads are increased on both sides when compared to the pre-operative hip disease group but there is still a significant difference between the two hips when the group is considered as a whole (Fig. 5.1). The increase in load may be a result of the increased walking velocity seen post-operatively (Paul, 1970). The lower loads across the replaced hip have to be attributed to the lower muscle forces seen in the abductor and extensor muscle groups (Chapter 4, Table 4.2b). Indeed, the most notable feature of this group is the failure of recovery in MHX (the abductor-adductor moment) at the replaced hip joints and the associated low abductor muscle forces (Fig. 5.2). As mentioned in the introduction to this section pain can be discounted as a cause of reduced muscle action in these patients, all of whom had a successful result from hip arthroplasty at one year. It is therefore necessary to consider the effects of the reconstructed anatomy upon hip muscle function.

Two types of hip arthroplasty were used and differences in joint loading can be seen between them (Fig. 5.3). In those patients treated with the C.A.D. Müller device the differences in peak hip joint loads is small reflecting the

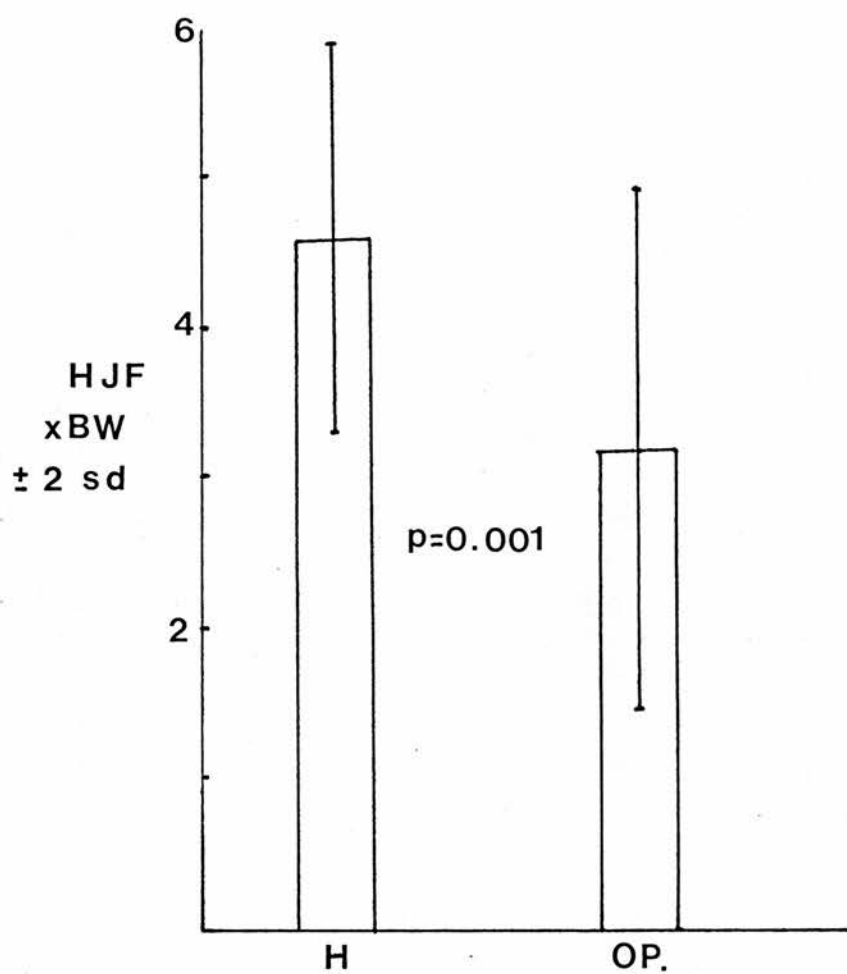


Fig. 5.1 Peak Hip Joint Loads After Hip Joint Replacement.

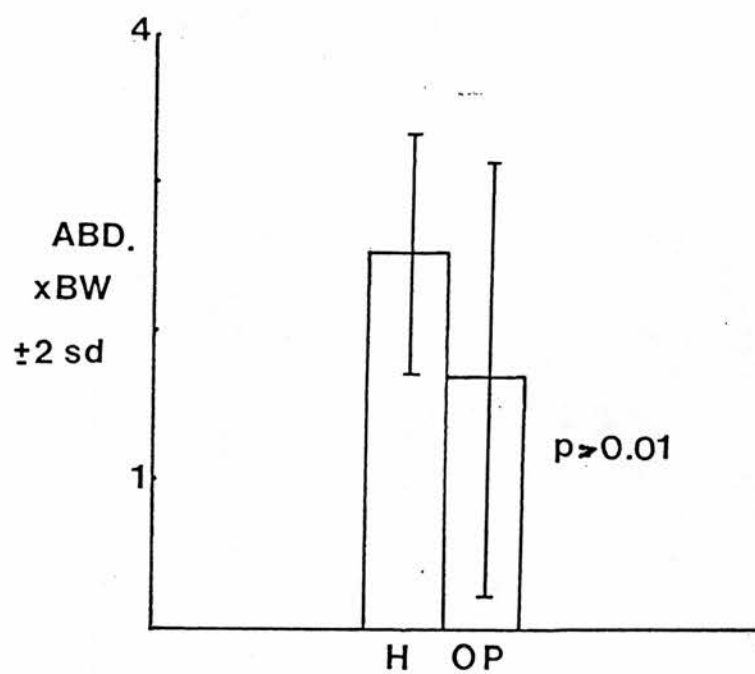


Fig. 5.2 Peak Hip Abductor Muscle Force
after Hip Joint Replacement.

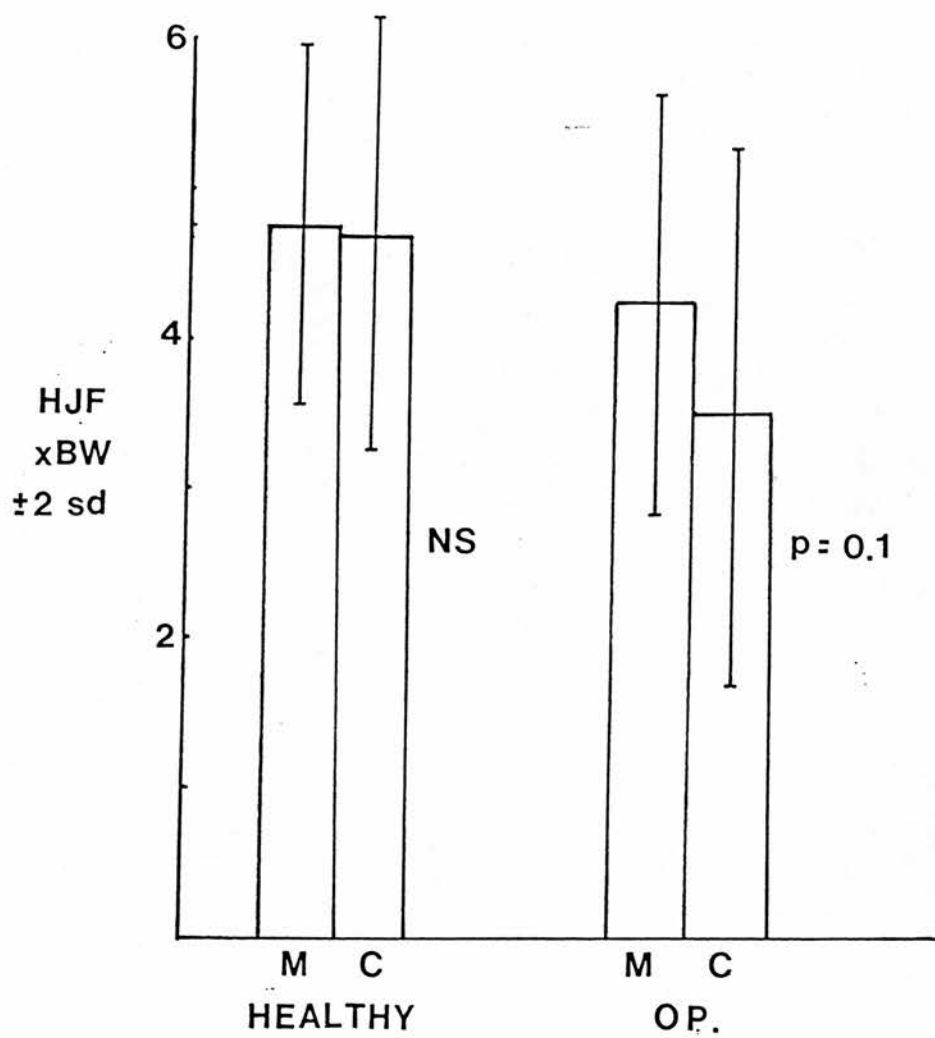


Fig. 5.3 Hip Joint Loads with regard to arthroplasty used
M = C.A.D. Müller. C = Charnley.

small difference in the abductor muscle force. This may be the result of the more anatomical reconstruction of the replaced hip joint in this group (Chapter 4, Fig. 4.64). In the patients treated with the Charnley device there is a greater difference between the hip joint loads; that at the replaced joint being lower than in the Müller group, whilst that at the healthy hip being of similar magnitude. This reflects the differences in abductor muscle force (Fig. 5.4). Indeed, in the Charnley group there is no significant difference in the abductor muscle force at the affected hip joint between the pre- and post-operative groups (pre-operatively 1.38 ± 1.72 xBW; post-operatively 1.32 ± 1.24 xBW).

The anatomy of the two groups is different as evidenced by Figure 4.64. The major feature post-operatively in the Charnley group is the medialisation of the hip joint centre with respect to the pelvis. Theoretically this will require less abductor muscle force to maintain equilibrium and will therefore lower joint force. Additionally, the method of re-attachment of the greater trochanter can increase the abductor muscle lever arm and this is seen in the Charnley group (Chapter 4, Table 4.9). This further reduces the requirement for abductor muscle action to maintain equilibrium. At this stage it should be noted that all greater trochanters had united at the time of the one year post-operative test.

If the mechanical explanation for the differing muscle force actions is complete one would expect to see curves of abductor moments, abductor muscle forces, and

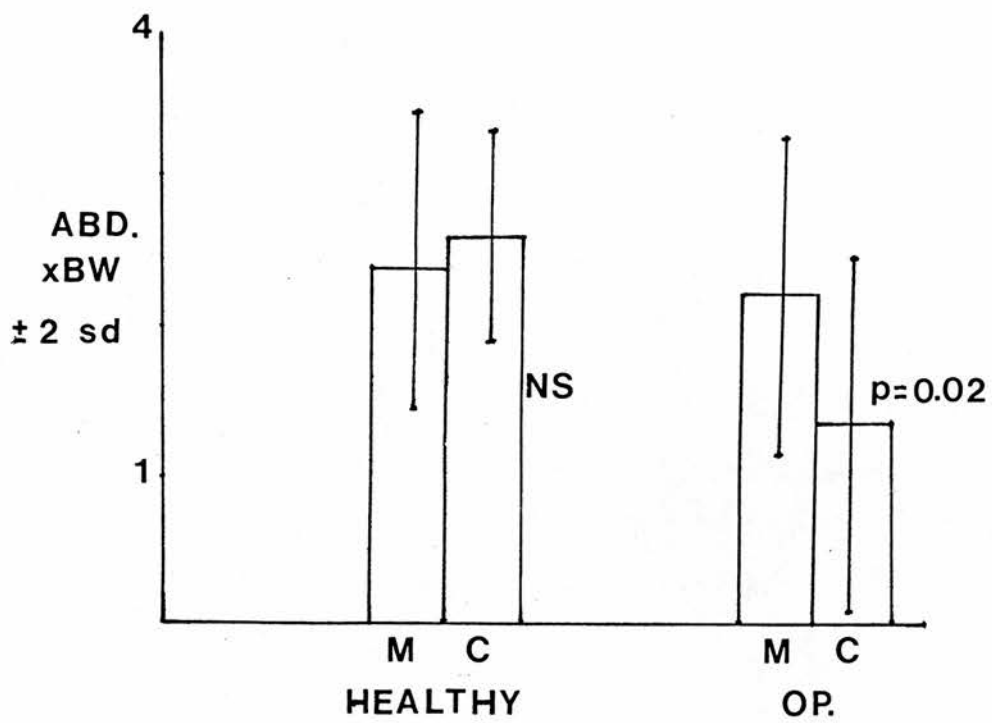


Fig. 5.4 Hip Abductor Muscle Force with regard to Arthroplasty Used
M = C.A.D. Müller. C = Charnley.

joint forces at the replaced hip reduced in value but of normal pattern in the Charnley group, and of similar magnitude to those for the contralateral hip, but again of normal pattern for the C.A.D. Müller group. This is the case in some patients but not in the majority where the pattern remains abnormal at the replaced hip, and, in some, these abnormalities are present at the healthy hip (Figs. 4.42, 4.43, 4.48 and 4.49). In the absence of pain three explanations may be considered for this observation. Firstly, there may be a persistence of the pre-operative gait pattern in the form of a habit. If pain is the major reason for the pre-operative gait this explanation is difficult to accept although impossible to disprove. Secondly, after hip arthroplasty with removal of the capsule and subchondral bone, proprioceptive impulses from the hip joint would be expected to be absent or reduced. This would result in a coarse control of muscular action and would be manifest as an abnormal pattern. It should be possible to test proprioception at the hip joint and the results of such work would be of considerable interest. Thirdly, the fixation of the prosthesis to bone, although adequate for function, does allow micro-movements. If proprioception is not impaired it is possible that such "instability" might result in the muscles being prevented from exerting their usual action.

3. Knee Joints

That the reduced muscle forces are not entirely due to anatomical factors is further suggested by a consideration of knee movements. The ipsilateral knees fail to regain

a normal pattern and exhibit a limited and fixed range stance phase suggestive of the need to reduce the requirement for moments about the hip joint. This could be partially the result of habit or proprioceptive loss at the replaced joint, but it is typical of the pattern seen where there is limitation of extension at the hip, e.g. hip arthrodesis (Gore et al, 1975).

Hip extension is not commonly measured clinically, and indeed initially was not in this work, although several patients were noted to have flexion contractures post-operatively. A later review of the patients confirmed that all but two patients (Chapter 2, Table 2.1) had no extension measurable at the hip joint. Those that did had been treated with a Charnley arthroplasty and had 10° of measurable extension.

However, extension is not the only hip movement which remains limited post-operatively. There is also failure of recovery of internal rotation at the replaced hip (Chapter 4, Table 4.1). Internal rotation of the femur on the pelvis during stance phase results from the horizontal rotation of the pelvis in its forward progression (Fig. 5.5). If this is not possible, or is limited, compensatory mechanisms will be required to facilitate ambulation. In part this will be achieved by the fixed, flexed position of the ipsilateral knee, since, in flexion it will permit transverse rotations.

The alignment of hip arthroplasties is usually such that it favours hip flexion (Fig. 5.6) and it might therefore be expected that excursion into extension would

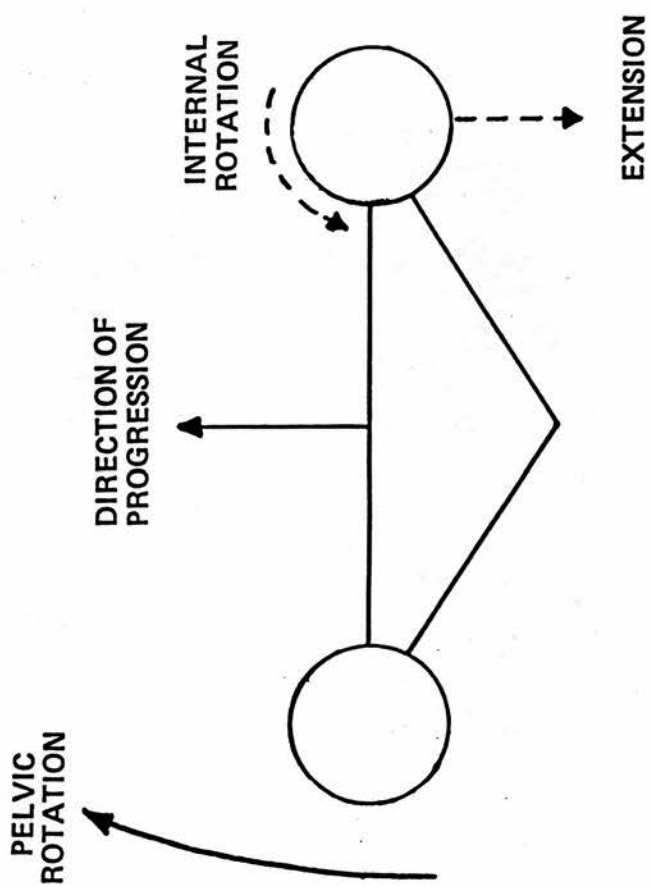
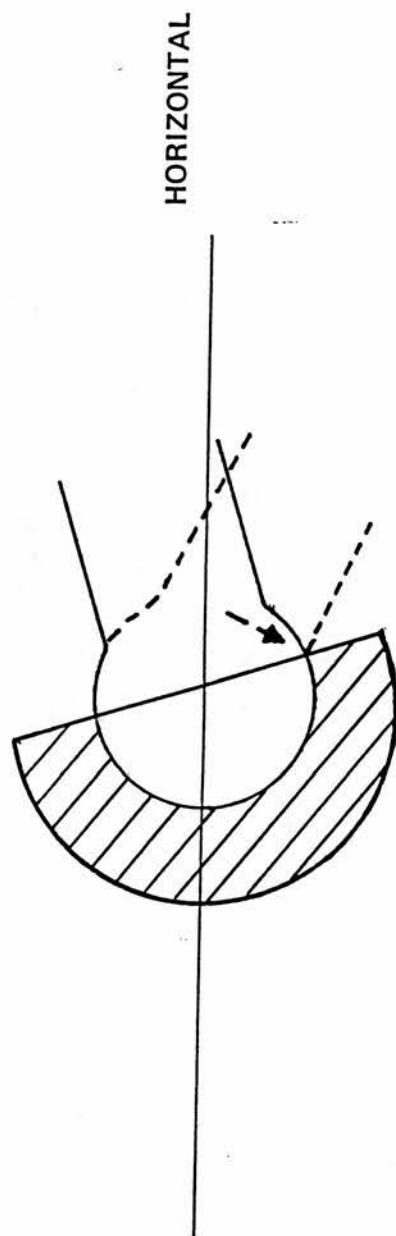


Fig. 5.5 Diagrammatic representation of pelvic and hip movements.

ANTERIOR



POSTERIOR

Fig. 5.6 Typical alignment of acetabular cup relative to the pelvis in a supine patient indicating impingement of neck on cup during extension.

predispose to anterior subluxation - a tendency which would be opposed by the hip musculature if any proprioceptive pathways were available.

Alternatively, scarring of the capsule and the failure to encourage hip extension post-operatively might be the cause of this finding. It would be of value to institute a programme of physiotherapy post-operatively which included the development of extension range and hip musculature to see if this produced any changes in the gait pattern and joint loads.

The lack of extension would also explain the persisting abnormality of increased post heel strike knee flexion in the contralateral knee joint where the knee flexion may, as suggested pre-operatively, be a result of increased lumbar lordosis used to increase the apparent extension range of the replaced hip and a similar pattern has been noted after hip fusion (Gore et al, 1975).

Alternatively, the fixed degree of flexion in the ipsilateral knee lowers the vertical position of the pelvis and may require increased knee flexion to compensate. Leg length discrepancies are small in this group and are unlikely to account for the degree of flexion seen at this knee joint (see Chapter 5.IV.2).

This mechanism is independent of the type of arthroplasty used and it has been shown that the load at the knee joints is also independent of the prosthesis type (Fig. 5.7). The load at the ipsilateral knee joint seems to depend upon the muscle function about the replaced hip joint and the limited internal rotation range, whilst that

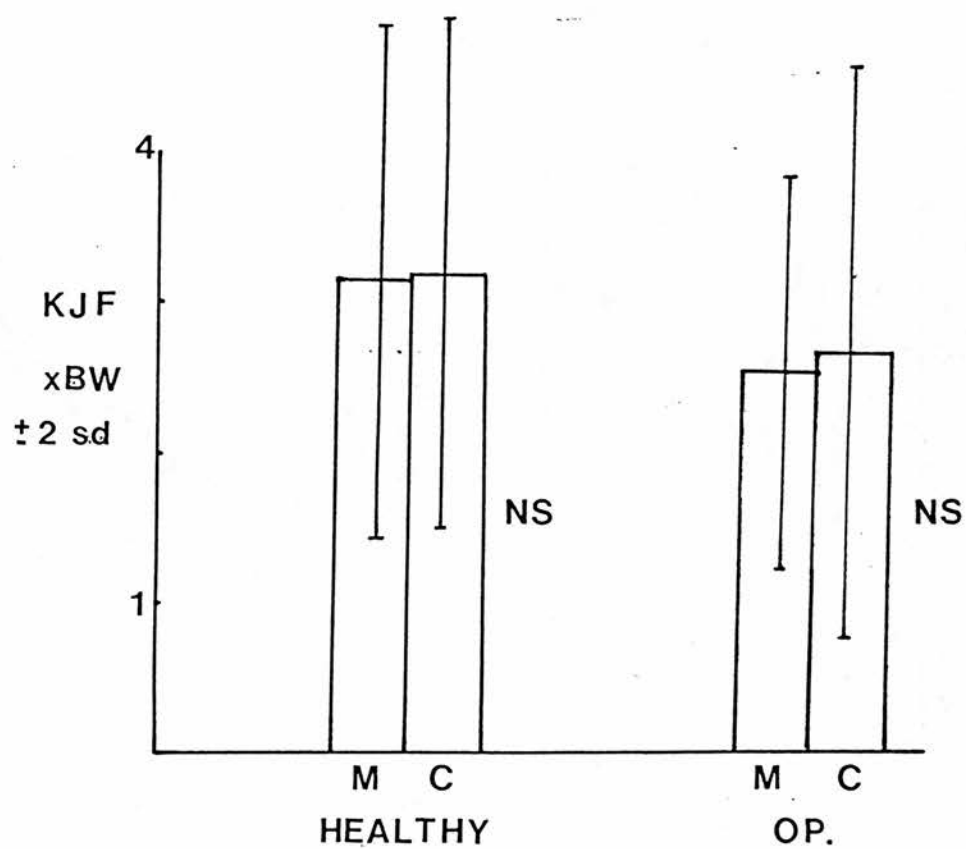


Fig. 5.7 Knee Joint Loads with regard to arthroplasty used
M = C.A.D. Müller. C = Charnley.

at the contralateral knee joint is a consequence of the increased knee flexion which may be secondary to limitation of extension in the hip arthroplasty.

Should the limitation of hip extension be confirmed as the cause of the knee movement pattern by further study and post-operative physiotherapy regimes prove to be ineffective, attention will have to be turned to the design of the arthroplasties. Nearly all current devices are simple ball and socket arrangements and, as such, are all subject to the limitations described here. The use of a bi- or tri-articular device might allow more extension but could cause problems by its very complexity. An alternative solution would be the use of an acetabular cup of less than a hemisphere to prevent impingement and allow a true extension range.

The compartmental loading pattern at the knees is very variable. For the group there is an increase in the amount of load borne by the lateral compartment of the ipsilateral knee. This would be expected as a result of the reduced abductor muscle force and moments which cause the load line to lie more laterally. Additionally, in a medialised joint with a medial shift in the position of the femoral shaft relative to the pelvis the ground force vector will lie more laterally within the knee joint. Persistent trunk sway of varying degrees might also be responsible although no patient had a noticeable limp one year post-operatively.

The contralateral knee joint loads are of a magnitude which is greater than that typically seen in normal

individuals during level walking (Morrison, 1967) and this is the more remarkable when it is considered that these patients are walking more slowly. If one accepts the work of Freeman (1974) and Chrisman (1982) linking joint load with degeneration, both prolonged walking with a diseased hip and walking after hip arthroplasty will subject the contralateral knee to an increased risk of degenerative change. When the articular cartilage has been put at risk, as in rheumatoid arthritis, such modest increases in load, as have been found in this work, might be sufficient to produce clinically apparent degeneration such as has been seen following hip arthroplasty in patients with rheumatoid arthritis (Hamblen, 1976).

The findings of Brand and Crowninshield (1980) that the use of a walking stick can reduce hip joint load and possibly knee joint load to a considerable extent suggests that the use of a stick on a long term basis after hip arthroplasty might be advantageous, especially in patients with rheumatoid arthritis. An investigation of the walking pattern and loads at all lower limb joints seen when using a stick would be of great interest.

4. Conclusions

i. After unilateral total hip joint replacement the patients walk with greater walking velocity than seen pre-operatively and the duration of stance phase on each limb is equal. However, asymmetry of hip and knee movements persist giving an abnormal pattern of gait.

ii. The gait pattern is felt to result from abductor and extensor muscle dysfunction at the replaced hip joint. The magnitude of the muscle forces seen can be explained on the basis of the reconstructed anatomy, but the abnormal pattern of the curves cannot be fully explained on the available data. The possibility of an abnormality of proprioception being involved has been raised.

iii. Limitation of extension and internal rotation at the hip joint following arthroplasty may be responsible for both limitation of knee movements at the ipsilateral knee and the increased knee flexion seen at the contralateral knee.

iv. The increased flexion at the contralateral knee joint results in an increase in knee joint force.

v. The movements and loads at the knee joints are independent of the type of arthroplasty used.

5.IV GIRDLESTONE EXCISION ARTHROPLASTY GROUP

1. Gait Pattern

This group of patients differs from the previous two groups in not having only one joint affected and in having only minor restrictions of passive movements. . Additionally, some patients used a walking stick thus further complicating the interpretation of the gait data. Pain was not a feature in this group.

The temporal and kinematic results for this group are very similar to those seen for the degenerative hip disease group. There was a 10 per cent reduction in the duration of stance phase on the affected limb, but as judged from the biomechanical tests there did not seem to be an irregular irregularity in the step lengths. There was a much reduced range of movement used at the affected hip during walking and the knee beneath this hip exhibited virtually no flexion during the period of single support. In the contralateral limb there was a great increase in the post heel strike range of knee flexion similar to that seen both in the degenerative hip disease and hip joint replacement groups.

As with the other two groups, abnormalities were seen in the moments about the hip joints. In particular there was a gross abnormality in the abductor adductor moment (MHX) at the hip joint with a much slower rise to peak value and an irregular plateau on the affected side.

2. Hip Joints

As suggested by the hip moments and as shown by analysis of the hip muscle forces there is clear evidence

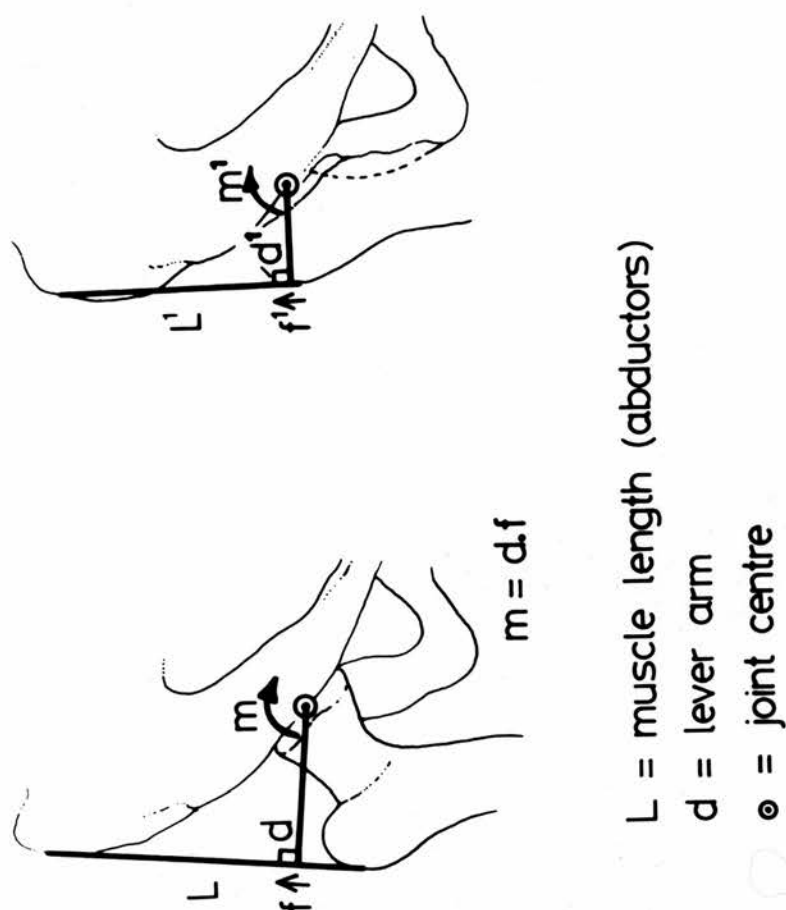


Fig. 5.8 Diagrammatic Representation of the Effect of Excision Arthroplasty on Hip Joint Anatomy.

of muscle dysfunction. From Figure 5.8 it can be seen that there is considerable distortion of the anatomy after Girdlestone excision arthroplasty of the hip and this results in shortening of the muscle lever arms as well as muscle fibre lengths. A combination of these factors will result in less muscle force being available to maintain equilibrium at the hip. There is therefore both structural and functional instability at the Girdlestone excision arthroplasty hip.

This structural and functional instability is reflected in a reduced range of movement at the hip and in a very much reduced movement at the ipsilateral knee. The limited movements of the knee, combined with the trunk sway, demand less of the hip musculature by causing the load line to pass through or close to the hip joint.

Limited movements at the hip joint produce a situation akin to that seen in the other two groups with no extension at the hip joint during gait. For the same reasons discussed in the previous two sections this could be responsible for the increased post heel strike knee flexion in the contralateral knee. However, limb length discrepancy is a major consideration in this group (Table 4.3). It is not uncommon to see 4 cms of limb shortening. Although no correlation could be found between the amount of limb shortening and the degree of contralateral knee flexion, a simple geometrical calculation shows that, to maintain a level pelvis, increased knee flexion will be required from the contralateral knee.

3. Knee Joints

The knee joint loads reflect the knee movement patterns. On the ipsilateral side the knee joint load is low and on the contralateral side is higher. However, the difference in knee joint loading in this group is not as marked as might be expected from a consideration of the knee movements. This is probably a result of the use of sticks by some patients, load being transmitted through the sticks rather than through the lower limbs. This could be more properly assessed by the use of an instrumented walking stick but such a device was not available at the time these tests were carried out.

The compartmental loading patterns at the knee for this group showed a consistent pattern with a clear increase in the lateral compartment loading in the ipsilateral knee. As mentioned above, the use of the GS limb is such that the load line will be brought more lateral to be near to the centre of rotation of the GS hip and this would bring the load line to lie more laterally in the knee joint. This mechanism is enhanced by the trunk sway which is a prominent feature of the gait of these patients. In the contralateral knee joint the load line passes through the medial joint compartment but the magnitude of the load is increased. This again raises the possibility of the occurrence of degenerative change in the knee joint but this has not been noted in a review of a large number of these patients (Campbell et al, 1978). This may be because of the slow walking velocity of this group of patients and may also be attributable to the fact

that the majority have to make use of a walking aid.

In view of these findings and since many patients who have to undergo this procedure have gross impairment of their walking ability it would be of great interest to institute a rehabilitation regime of physiotherapy to develop the hip musculature in an attempt to improve gait. It should be noted that none of the patients showed "telescoping" and therefore the measured leg length discrepancy was also that which pertained during walking.

4. Conclusions

i. After Girdlestone excision arthroplasty of the hip joint for failed total hip replacement any abnormal gait is seen similar to that seen in patients with gross hip disease. There is a 10 per cent reduction in the duration of stance phase of the limb of the affected side together with a reduction in the range of movement used at the affected hip joint. The ipsilateral knee joint shows very little movement indeed during stance phase but there is an increased range of movement at the contralateral knee. The velocity of walking is slow.

ii. This gait pattern results from impairment of muscle function about the hip brought about by the gross anatomical distortion and also by the resulting instability.

iii. There is an increase in the loading of the lateral joint compartment in the ipsilateral knee compared to normal subjects but a higher load across the contralateral knee joint.

5.V OTHER CONSIDERATIONS

1. Loads and Movements at the Healthy Hips

Although the peak loads seen at the healthy hips in all groups are of the same magnitude as those seen in normal subjects (Paul, 1967), movement patterns and the pattern of the load curves are abnormal.

The limitation of extension and internal rotation at the affected and replaced hips has been noted. During gait, transverse rotations of the pelvis are needed to facilitate progression (Fig. 5.5). These limitations of movement at one hip will limit the amount of transverse rotation possible when standing on the affected limb, particularly if the knee joint is not flexed. There will therefore have to be modification of movements at the healthy hip to allow a reasonable stride length to be maintained. Similarly, inability of the abductors to function fully will alter the amount of pelvic tilt available for use and will again affect movements at the healthy hip.

Further, axial rotations of the limb segments are essential to locomotion (Inman, 1981). Because of the technical difficulties involved these have not been measured in this study. Also, because of the difficulty in estimating the phasic activity in the hip rotator muscles they have not been considered in the analysis. The moments about the vertical axis (MHY) have not been presented because of the great errors possible in their estimation. However, the magnitude is reduced in the pre-operative hip disease group and increased post-operatively although, in keeping with the findings of other hip moments, they do not reach

normal magnitude. This suggests that limb segment rotation remains abnormal in the affected limb after hip replacement. If this is so it will have a major effect upon pelvic movements and thus upon the movements of the healthy limb.

It is therefore desirable to study limb segment rotations during gait in this group of patients and different techniques of gait analysis will be necessary for this study.

2. Anatomical Variables

It was possible to derive only limited information from the correlation and multiple linear regression studies concerning the anatomical variables.

In the hip disease group where it has been argued that pain is an important factor in relation to the gait, the loads at the affected hip were strongly related to the vertical position of the joint centre (Table 4.6a), the higher this was in the pelvis the lower was the joint force. The neck length of the affected hip (N.L.) also strongly influenced this load, the latter being lower the shorter the neck length. This suggests that the joint forces are lower in the more radiologically severe hips. However, these anatomical features may not be causative since such patients usually walk more slowly and will have lower joint forces by virtue of that fact (Paul, 1970; Andriacchi, 1977).

However, the length of the neck of the affected hip also explains much of the variability in the loads

at the contralateral hip and knee - the shorter the neck length, the higher the loads in these joints. This may again reflect the severity of the disease.

Post-operatively it was certainly not possible to confirm the findings of Johnston et al (1979) with regard to the positioning of the components of the arthroplasty. They suggested that :-

i. in order to reduce loads at the hip joint the joint centre should be moved as far medially, inferiorly and anteriorly as is anatomically possible;

ii. moving the greater trochanter laterally is mechanically desirable but less significant;

iii. a short prosthetic neck length and femoral shaft - prosthetic neck angle of 130° appeared to be optimum, the joint contact force being increased with higher angles although not being affected with neck length.

The results presented in this work indicate that a high and not a low position of the acetabular cup favours a low hip joint force and it also indicates that the greater the horizontal distance between the hip joint centre and the tip of the greater trochanter (G.T.Z.) the lower is the load across the contralateral knee joint. Although the position of the greater trochanter can be varied in the patients with Charnley arthroplasty, in those who have undergone Müller arthroplasty it is the valgus position of the femoral neck which determines the horizontal distance between the greater trochanter and

hip joint centre. The results of the linear regression analysis for the small Müller group suggests very great importance of this feature with regard to the knee joint load although it does not figure importantly when the Charnley group is investigated. When the entire hip joint replacement group is analysed the most notable feature is the lack of any association between the design of the prosthesis used and the loads at the knee joints. Since there is no statistically significant difference between the loads at the knee joint (Fig. 5.7) it is likely that these statistical results reflect the importance of the measurement G.T.Z. in two or three individuals within the Müller group.

A much larger series of patients treated with both types of arthroplasty requires to be studied and analysed statistically in an attempt to obtain meaningful information. This would be of very great value to the orthopaedic surgeon who, by his positioning of the arthroplasty, has the power to influence hip joint mechanics and possibly, the pattern of lower limb joint loading.

3. Sources of Error

A full error analysis of the gait analysis system used has not been performed. This system has superseded the cine film system used by Paul (1967) where a total error of 24 per cent was estimated comprising measurement errors and errors introduced by the making of assumptions such as muscle lines of action. An effective error rate of some 10 per cent was proposed. The T.V. system eliminates phasing errors and should reduce the total

errors to a small degree. However, in this study, an additional source of error has been the measurement of anatomical structures from x-rays. Whilst attempts were made to obtain standard radiographs in every case this was not always achieved. Weisman and his colleagues in 1978 estimated that the accuracy of measurements from x-rays was of the order of ± 0.5 cms. In this study, comparisons between measurements made from x-ray were done on a proportionate basis, to the standard of a known anatomical dimension. Hence the errors would not be of the same order as that found by Weisman.

The overall error rate is therefore probably the order of 15 per cent. Improvements in this could be made by a better procedure for obtaining radiographs and perhaps, by using a system of computerised measurement of the x-rays. Improvements in the data acquisition system would include the use of a stroboscope to "freeze" images of the markers (indeed this has already been introduced) and radiological checks of the marker positions, although this would introduce medical ethical considerations.

However, the error rate is typical of those encountered with non-invasive estimations of joint load and no major improvements will occur without a greater understanding of muscle function during gait and the introduction of a system for the measurement of muscle forces during walking.

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